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(54) **REMOVAL OF A-SCAN STREAKING ARTIFACT**

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G06T 7/00	(2006.01)
G06T 5/00	(2006.01)

(52) **U.S. Cl.**

CPC ***G06T 7/0012*** (2013.01); ***G06T 5/002*** (2013.01); *G06T 2207/10101* (2013.01); *G06T 2207/20012* (2013.01); *G06T 2207/30101* (2013.01)

(58) **Field of Classification Search**

CPC G06T 7/0012; G06T 2207/10101; G06T 2207/20012; G06T 2207/30101; G06T 5/002

See application file for complete search history.

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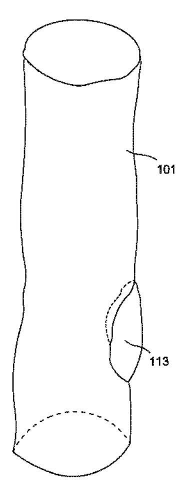
(57) **ABSTRACT**

This invention generally relates to the removal of streaking artifacts and periodic noise from tomographic images. The method comprises obtaining an A-scan from an imaging data set. The A-scan having a signal and the signal defining an amplitude. Noise specific to the A-scan is estimated. The amplitude of the A-scan is scaled based on its specific estimated noise floor. In another aspect, a plurality of A-scans is obtained from an imaging data set. Each of the plurality of A-scans has a signal and the signal defines an amplitude. Noise specific to each A-scan of the plurality of A-scans is estimated. Each A-scan of the plurality of A-scans is scaled by the A-scan's specific estimated noise floor.

16 Claims, 23 Drawing Sheets



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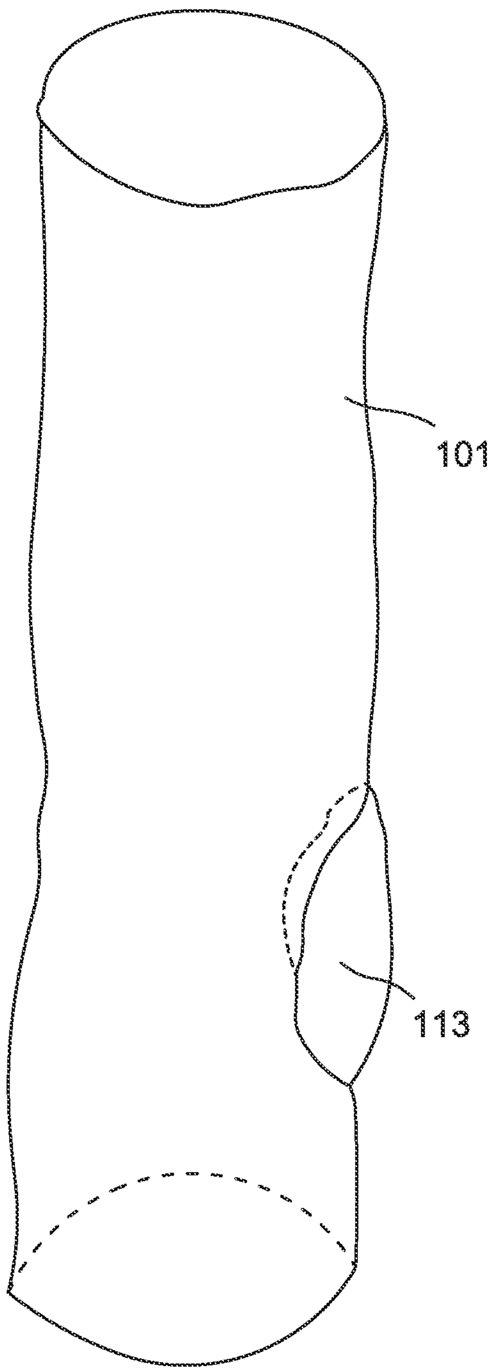


FIG.1

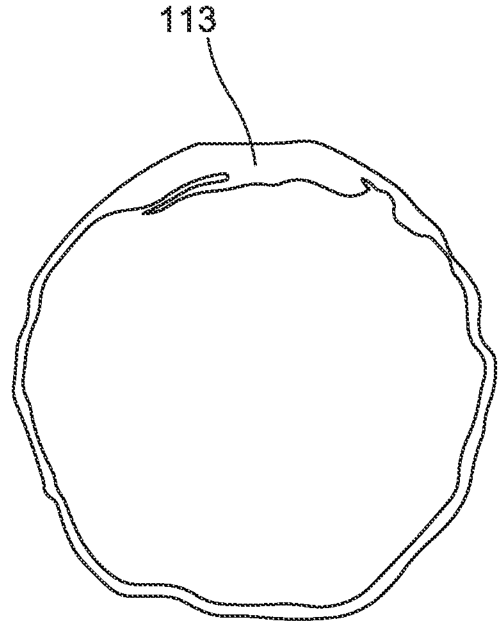


FIG.2

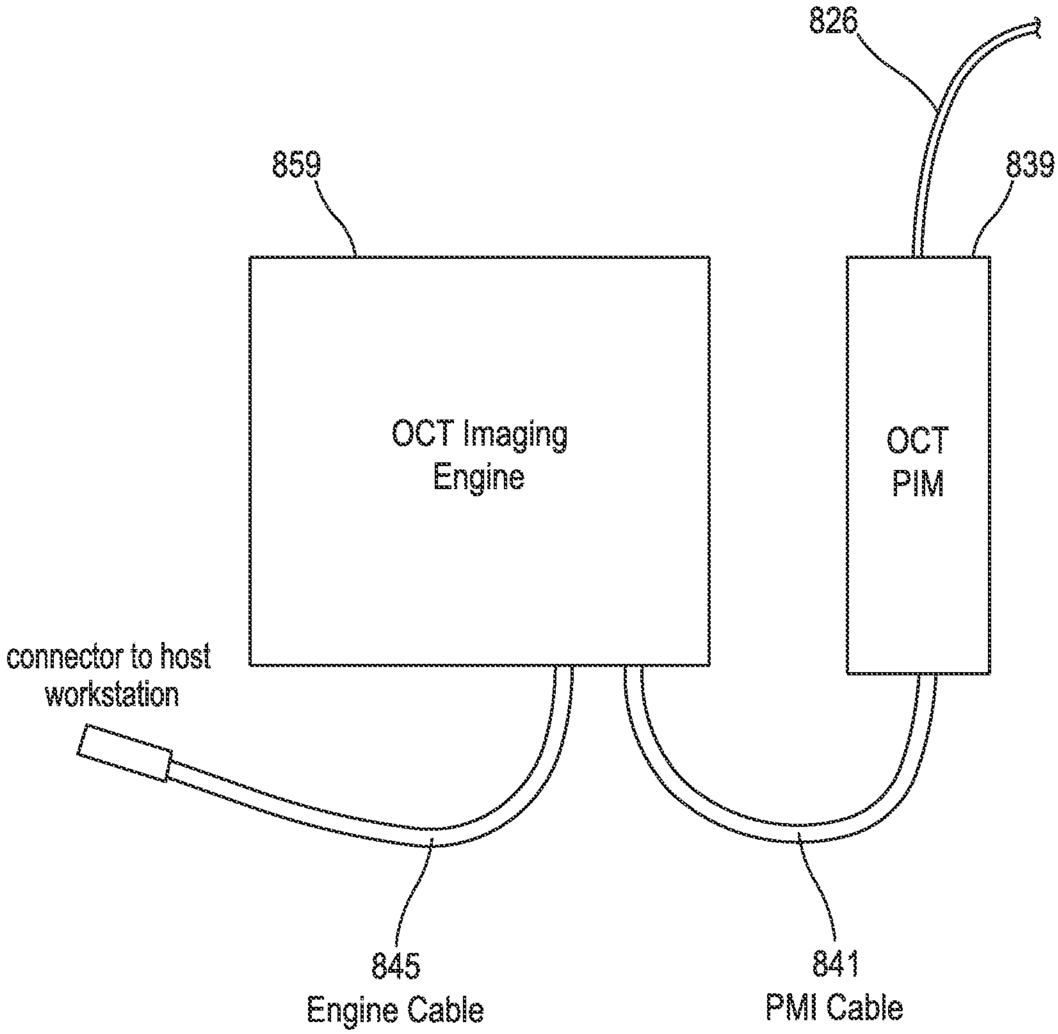
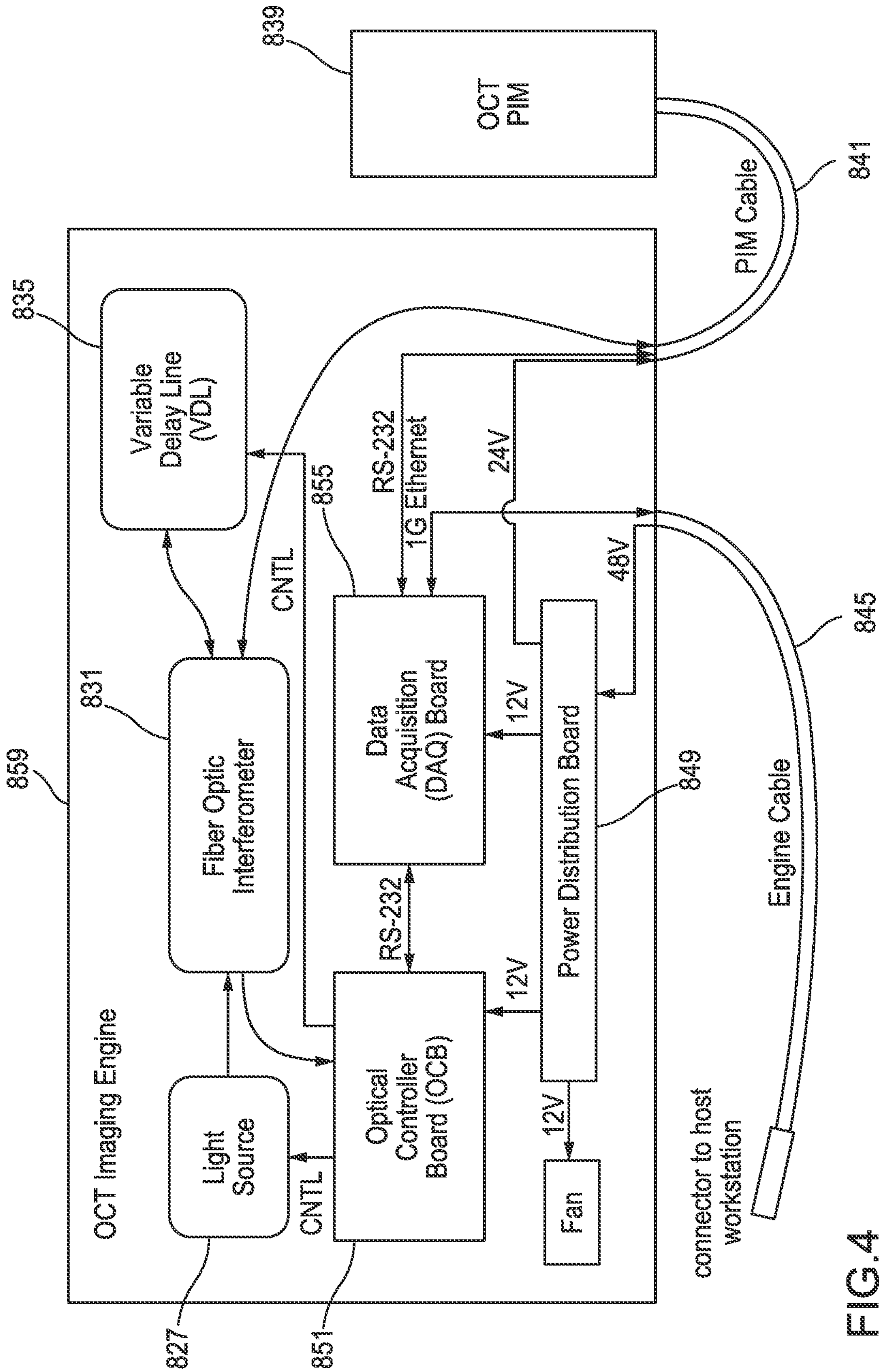

826
859
839
OCT Imaging
Engine
OCT
PIM
connector to host
workstation
845
Engine Cable
841
PMI Cable

FIG.3



839
OCT PIM
841
PIM Cable
835
Variable Delay Line (VDL)
859
OCT Imaging Engine
831
Fiber Optic Interferometer
855
RS-232
1G Ethernet
24V
CNTL
Data Acquisition (DAQ) Board
48V
845
12V
Power Distribution Board
849
Engine Cable
RS-232
12V
Light Source
Optical Controller Board (OCB)
CNTL
12V
Fan
connector to host workstation
827
851

FIG. 4

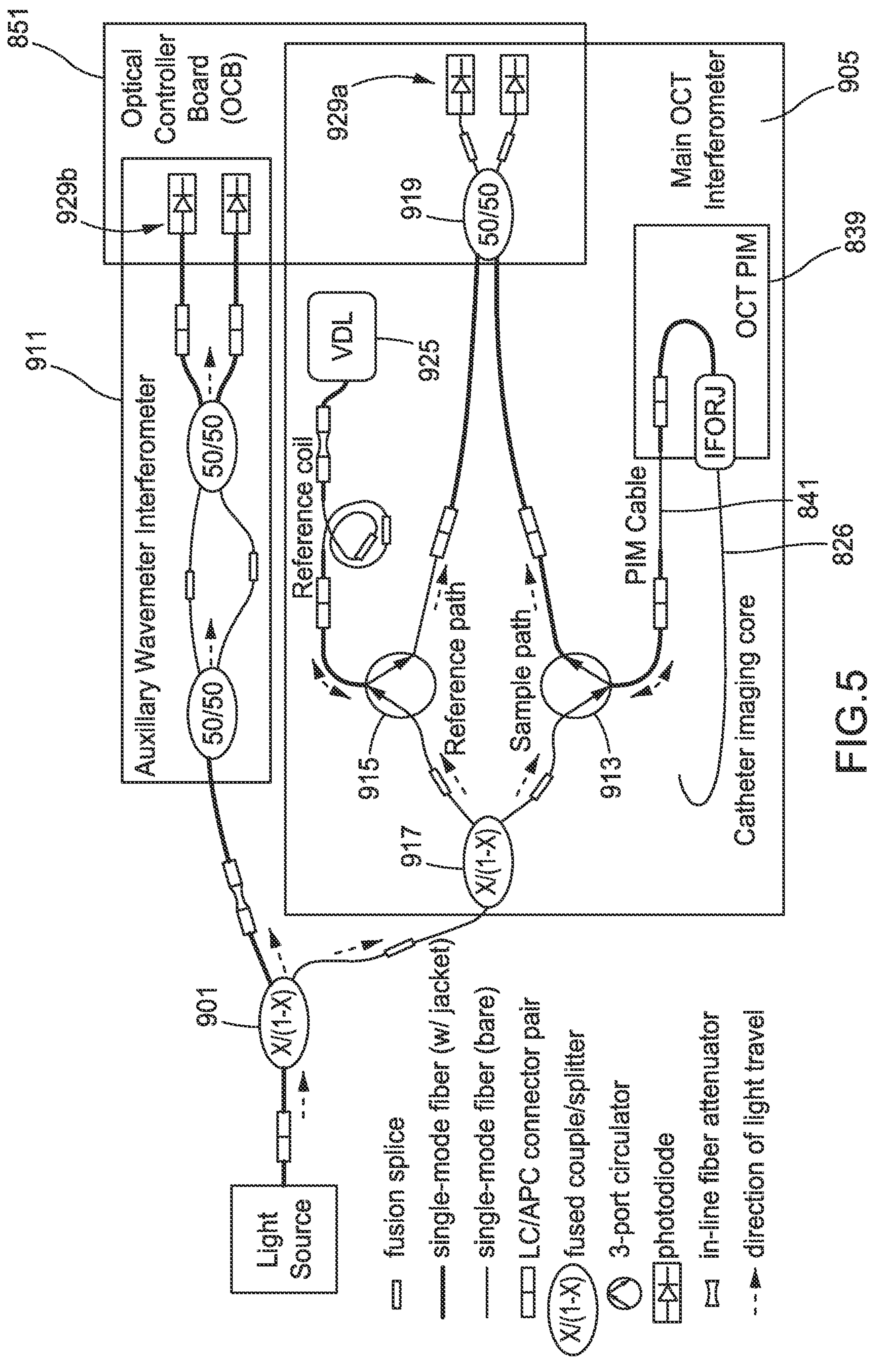

851
Optical Controller Board (OCB)
929b
929a
911
Auxiliary Wavemeter Interferometer
50/50
50/50
919
50/50
905
Main OCT Interferometer
VDL
925
839
OCT PIM
IFORJ
Reference coil
PIM Cable
841
826
Reference path
Sample path
Catheter imaging core
915
913
917
X/(1-X)
901
X/(1-X)
Light Source
fusion splice
single-mode fiber (w/ jacket)
single-mode fiber (bare)
LC/APC connector pair
X/(1-X) fused couple/splitter
3-port circulator
photodiode
in-line fiber attenuator
direction of light travel

FIG.5

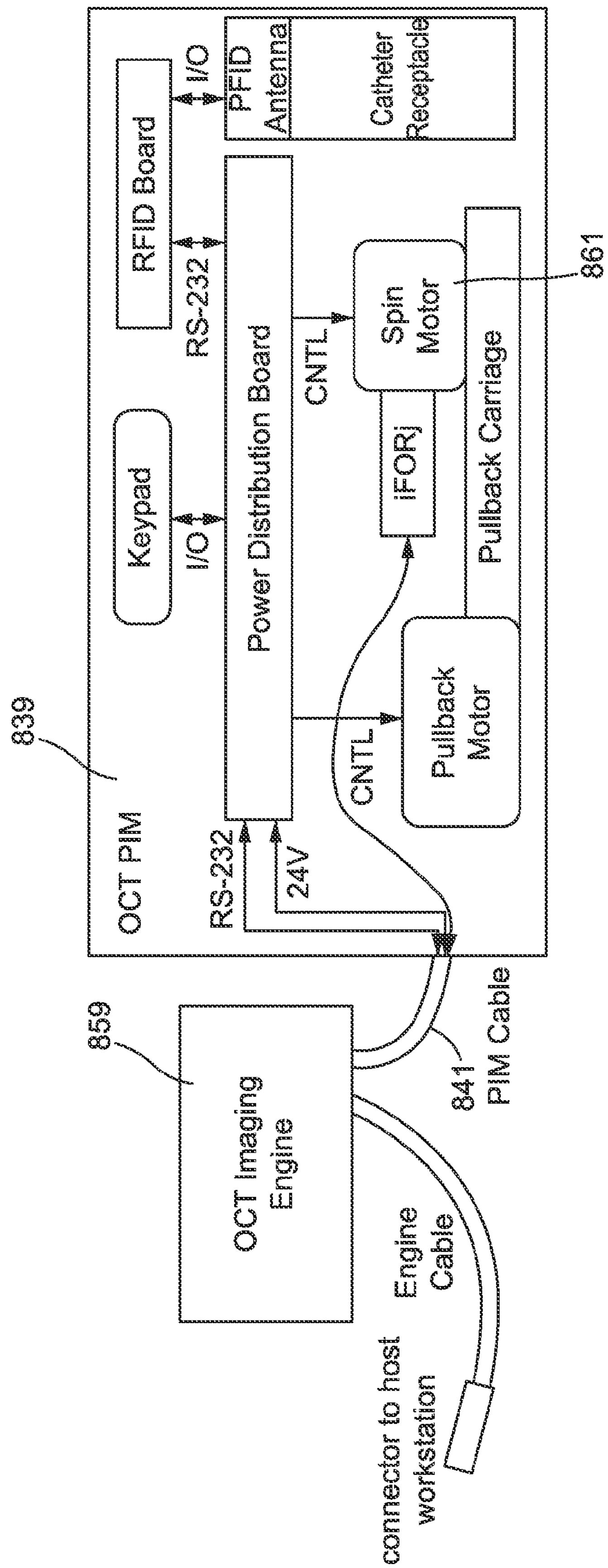

839
OCT PIM
Keypad
RFID Board
I/O
RS-232
I/O
PFID Antenna
Catheter Receptacle
Power Distribution Board
CNTL
Spin Motor
861
IFORJ
Pullback Carriage
CNTL
Pullback Motor
RS-232
24V
859
OCT Imaging Engine
841
PIM Cable
Engine Cable
connector to host workstation

FIG.6

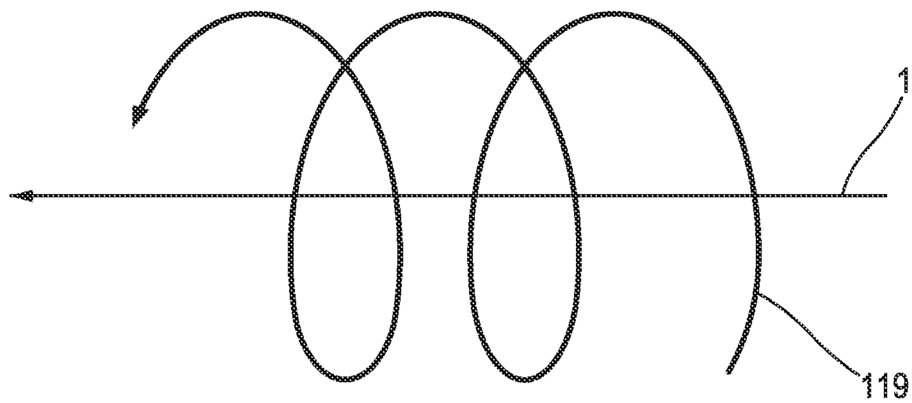


FIG.7

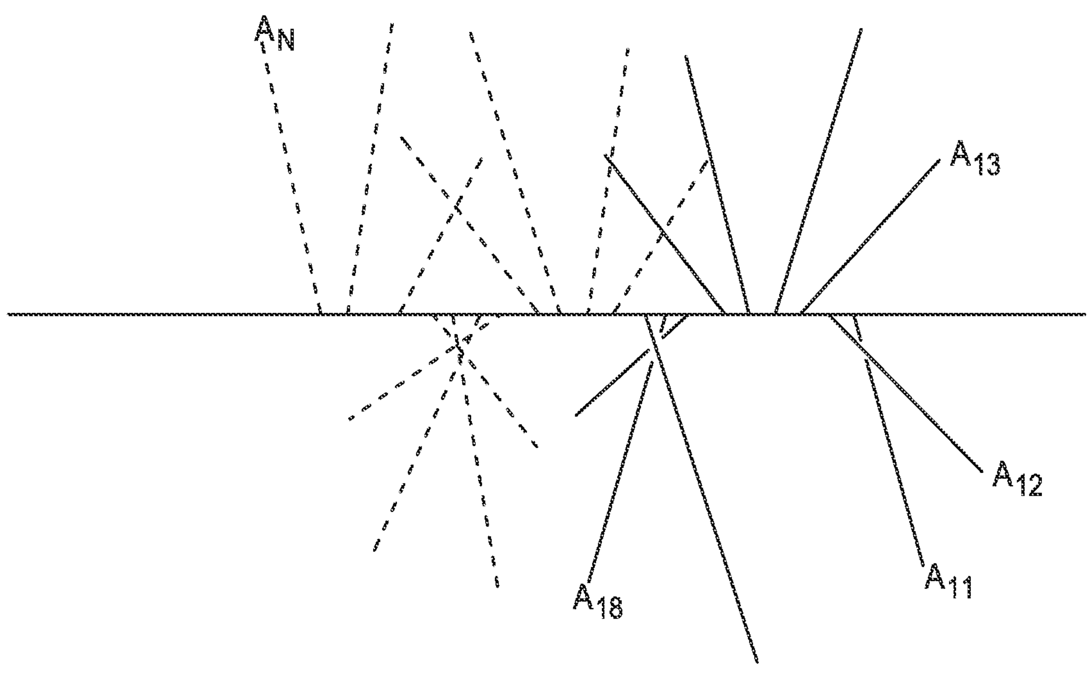


FIG.8

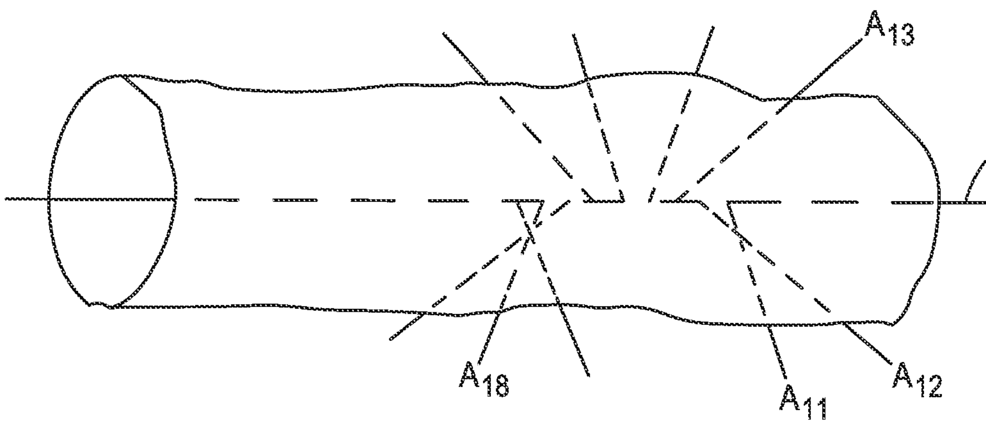


FIG.9

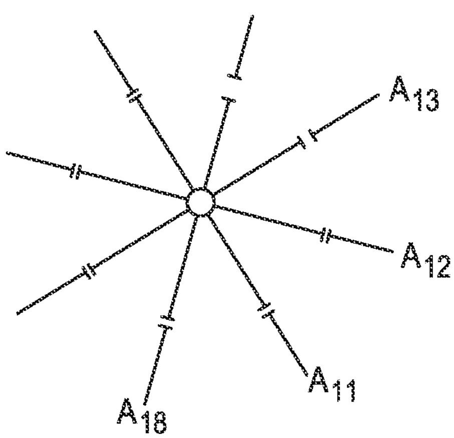


FIG.10

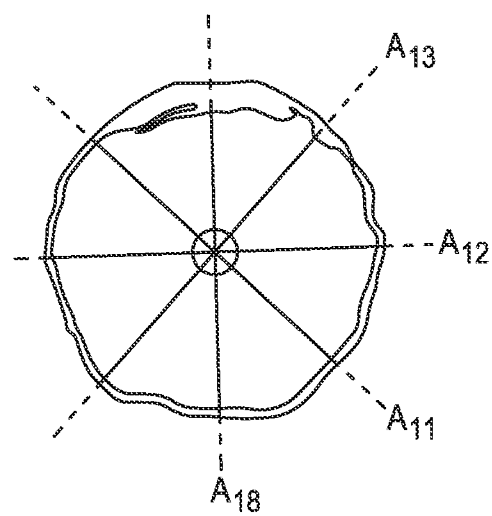


FIG.11

Figure 12

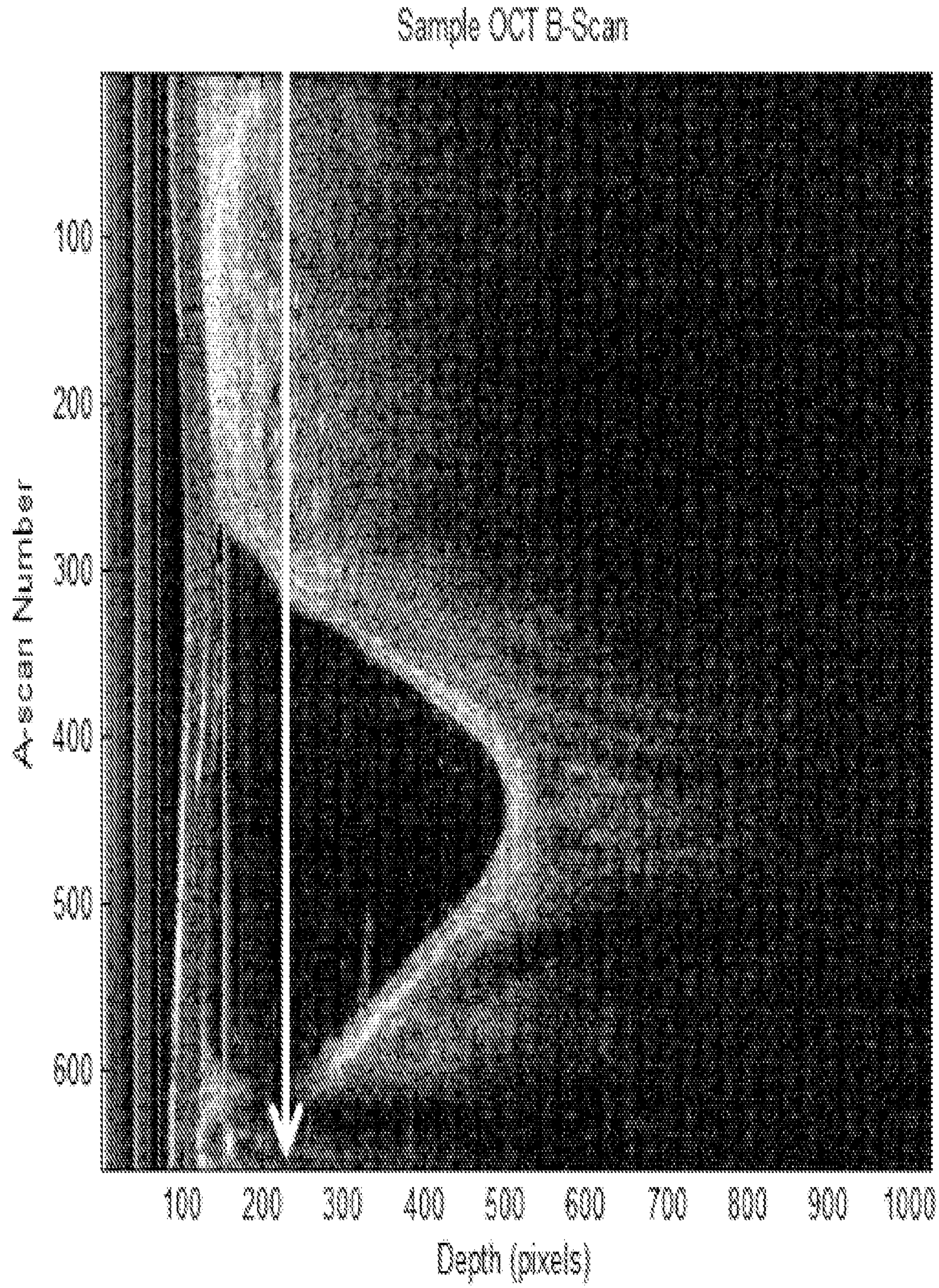
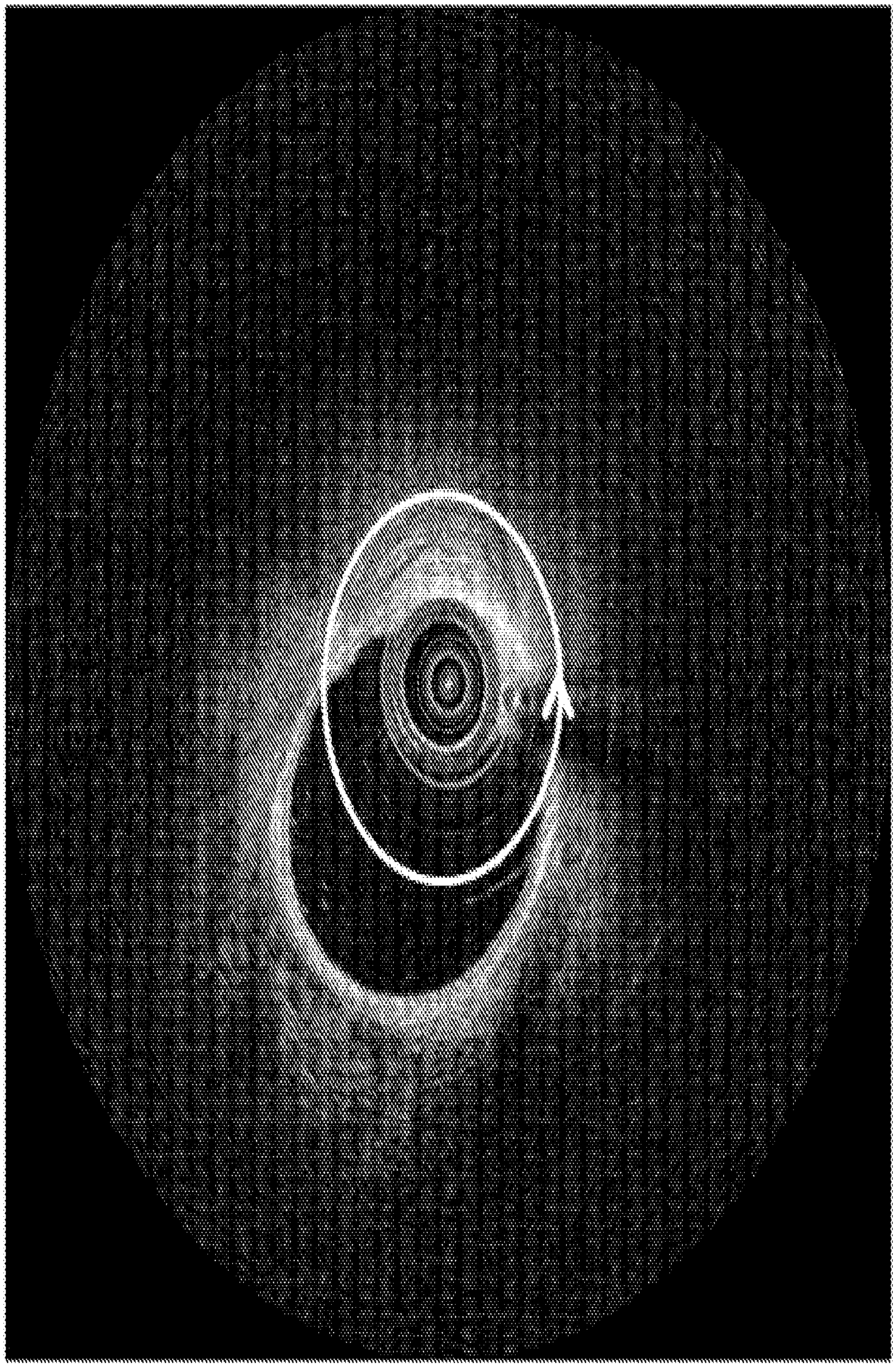

Sample OCT B-Scan
A-scan Number
100
200
300
400
500
600
100
200
300
400
500
600
700
800
900
1000
Depth (pixels)

Figure 13

Scan Converted Example



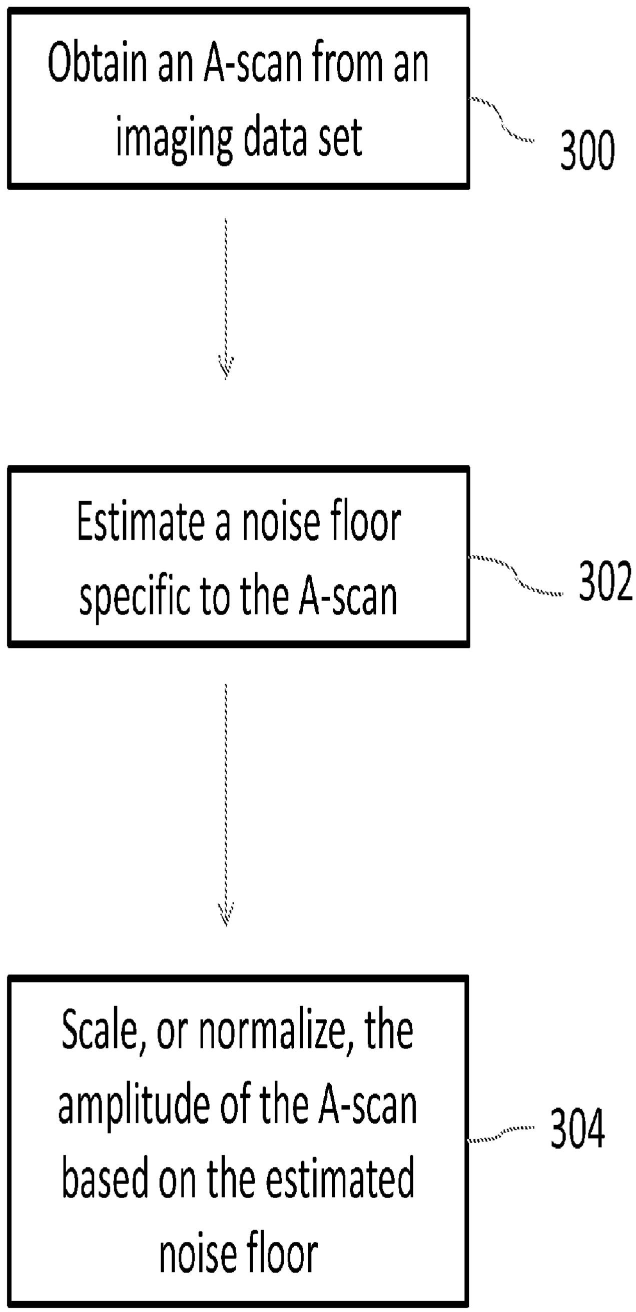

Obtain an A-scan from an imaging data set
300
Estimate a noise floor specific to the A-scan
302
Scale, or normalize, the amplitude of the A-scan based on the estimated noise floor
304

Figure 14

Figure 15

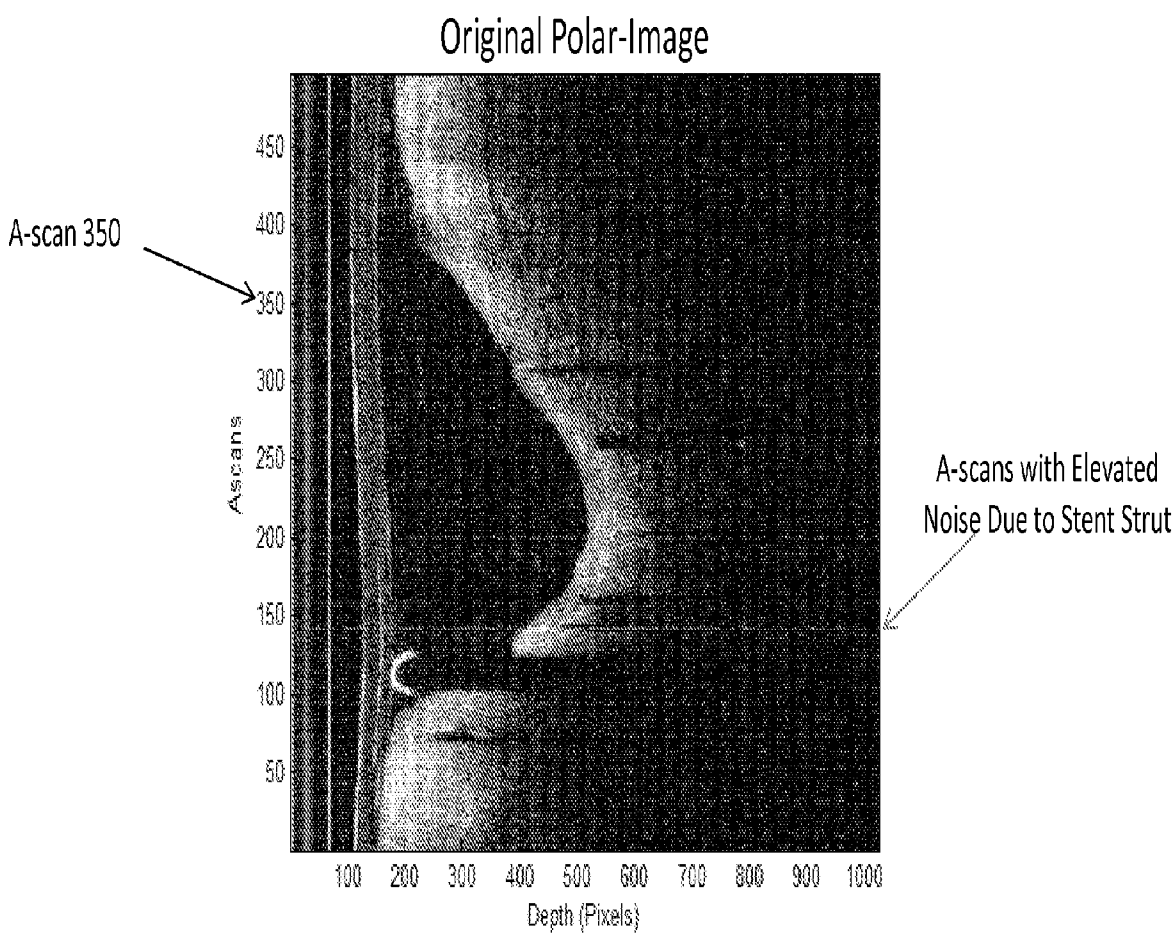

Original Polar-Image
A-scan 350
Ascans
450
400
350
300
250
200
150
100
50
100 200 300 400 500 600 700 800 900 1000
Depth (Pixels)
A-scans with Elevated Noise Due to Stent Strut

Figure 16

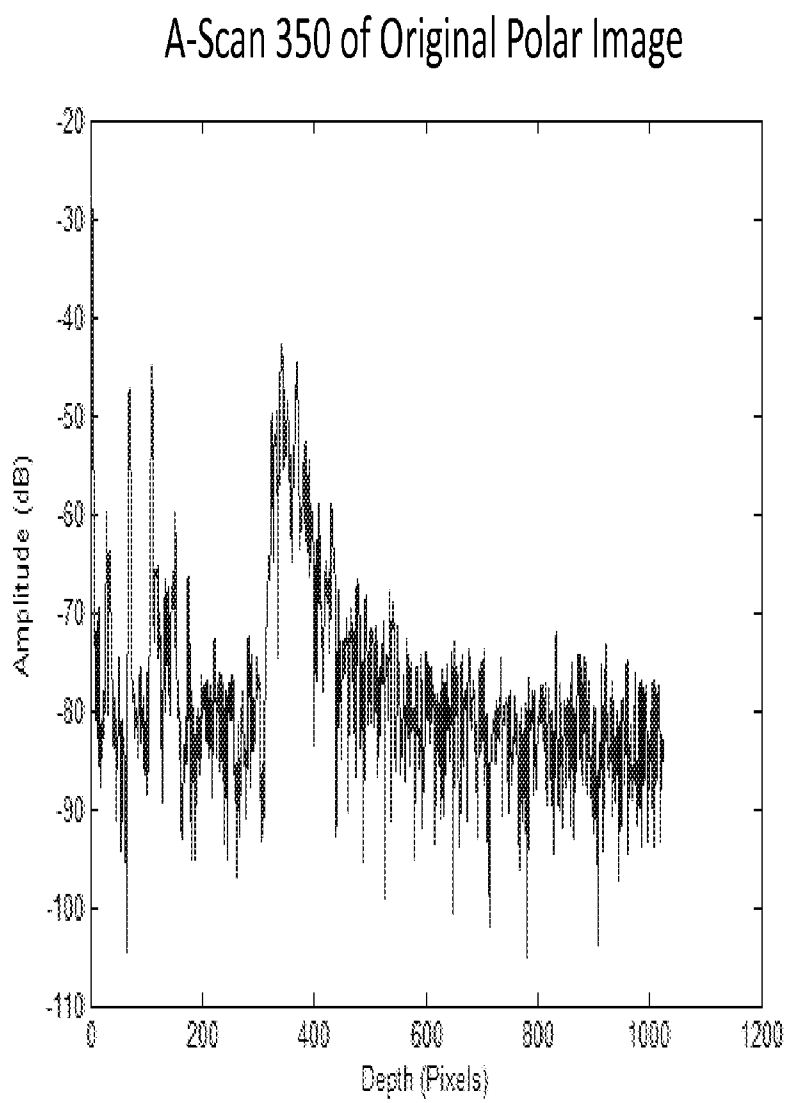

A-Scan 350 of Original Polar Image
-20
-30
-40
-50
-60
-70
-80
-90
-100
-110
Amplitude (dB)
0
200
400
600
800
1000
1200
Depth (Pixels)

Figure 17

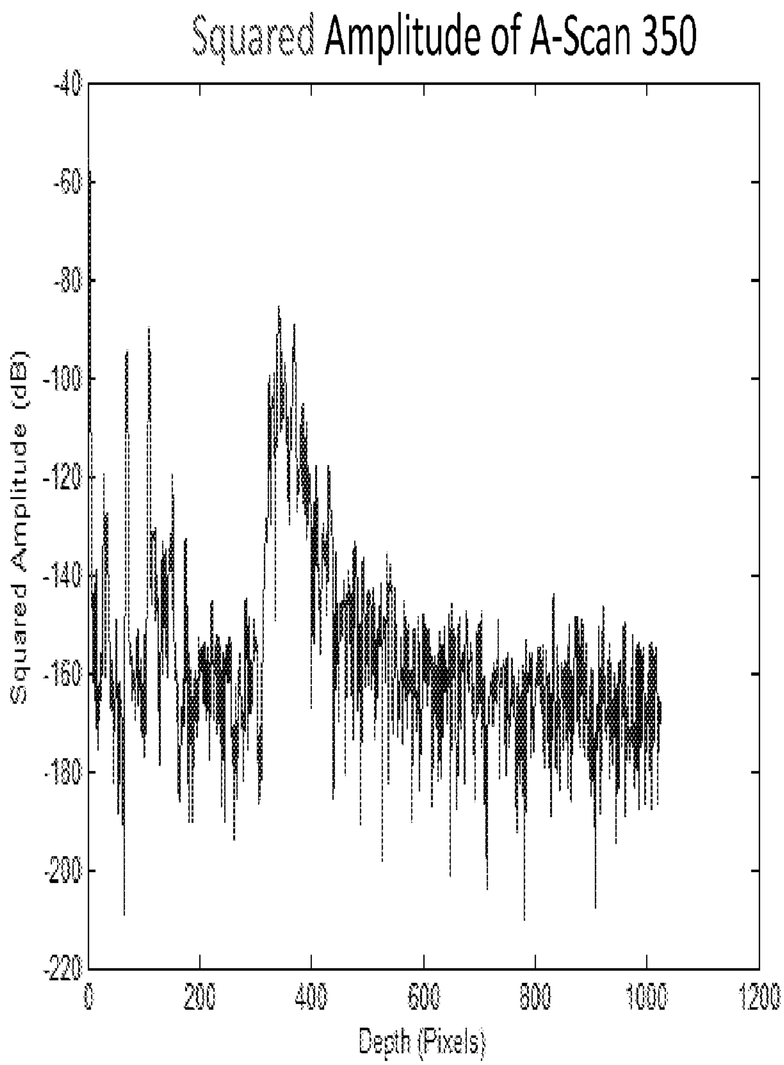

Squared Amplitude of A-Scan 350
Squared Amplitude (dB)
Depth (Pixels)
-40
-60
-80
-100
-120
-140
-160
-180
-200
-220
0
200
400
600
800
1000
1200

Figure 18

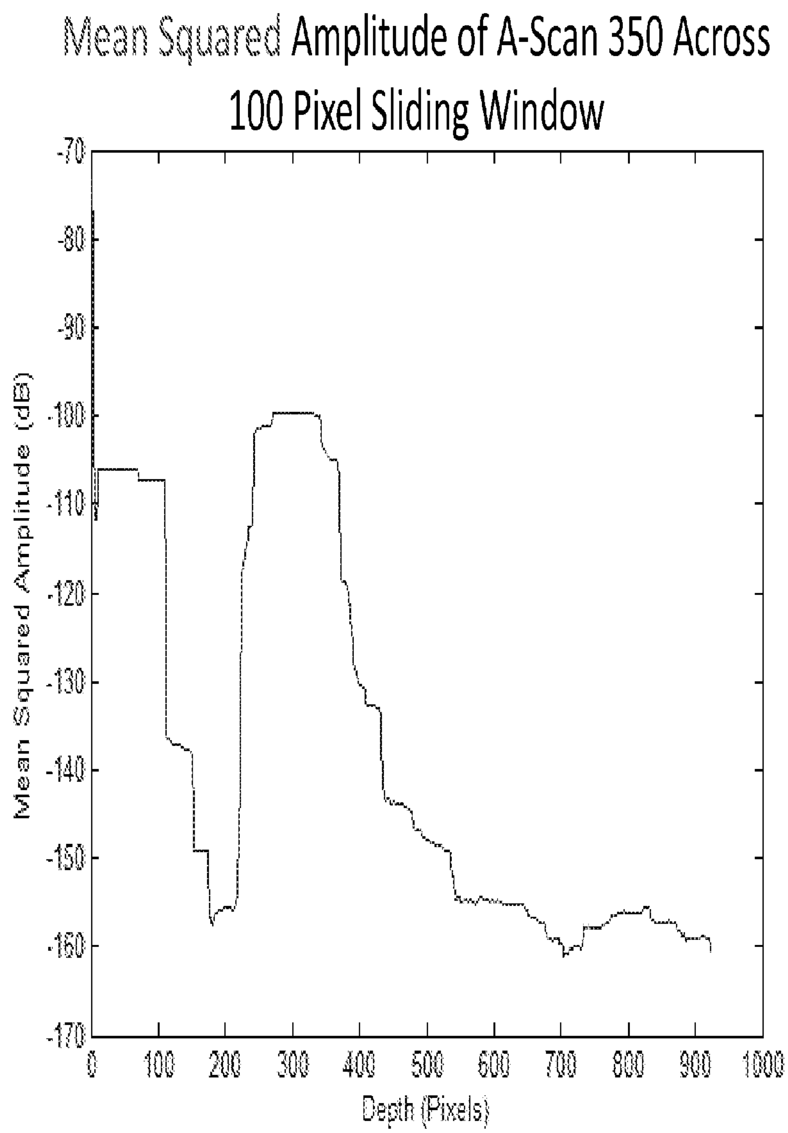

Mean Squared Amplitude of A-Scan 350 Across
100 Pixel Sliding Window
Mean Squared Amplitude (dB)
-70
-80
-90
-100
-110
-120
-130
-140
-150
-160
-170
0
100
200
300
400
500
600
700
800
900
1000
Depth (Pixels)

Figure 19

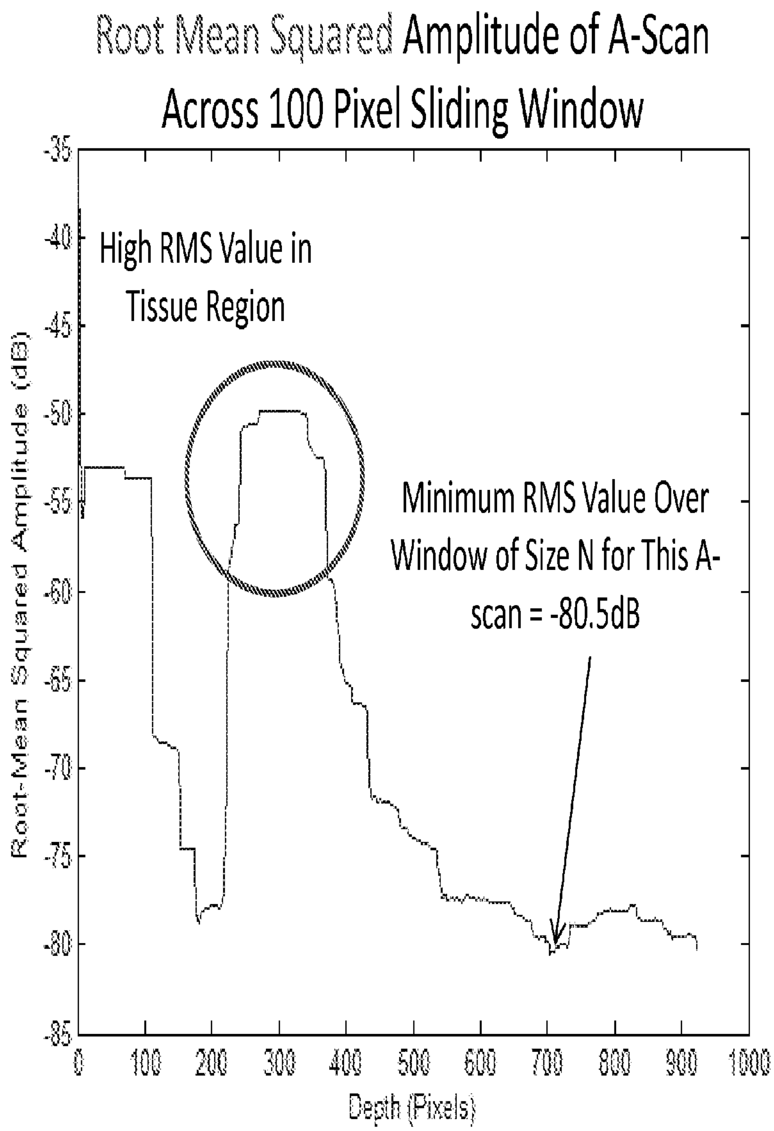

Root Mean Squared Amplitude of A-Scan
Across 100 Pixel Sliding Window
High RMS Value in
Tissue Region
Minimum RMS Value Over
Window of Size N for This A-
scan = -80.5dB
Root-Mean Squared Amplitude (dB)
Depth (Pixels)
-35
-40
-45
-50
-55
-60
-65
-70
-75
-80
-85
0
100
200
300
400
500
600
700
800
900
1000

Figure 20

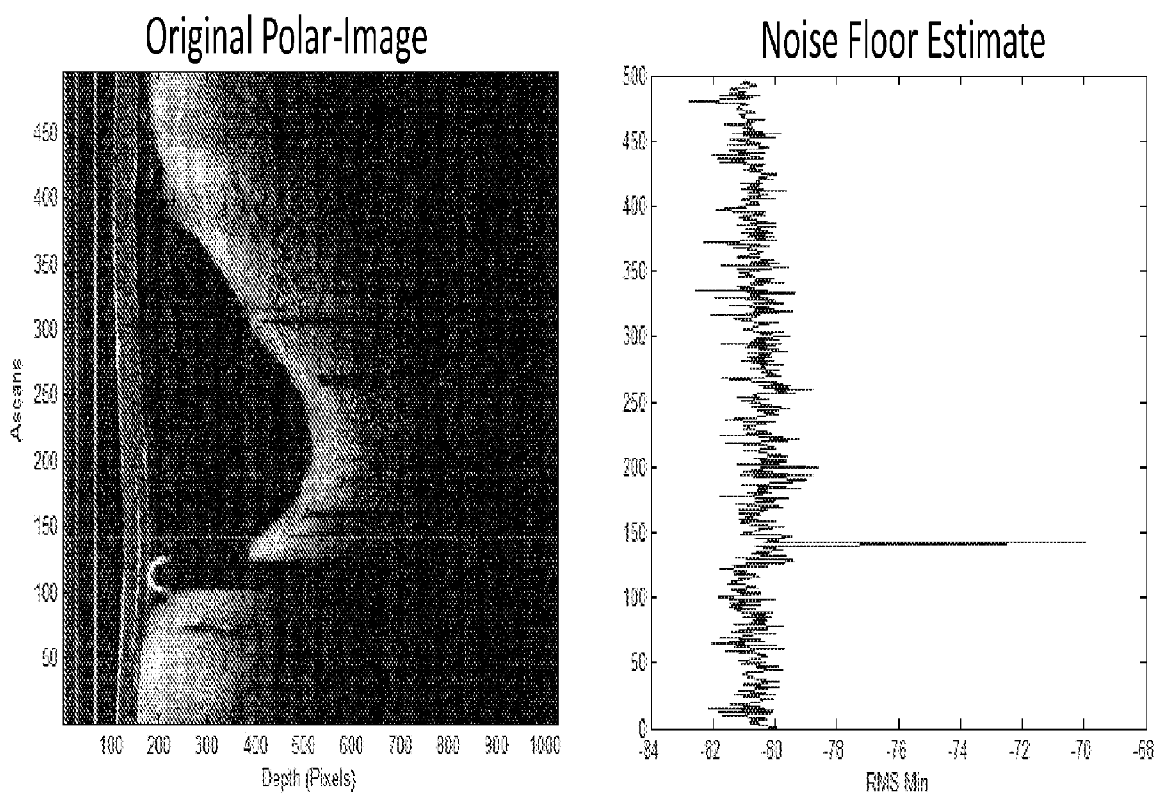

Original Polar-Image
Ascans
450
400
350
300
250
200
150
100
50
100 200 300 400 500 600 700 800 900 1000
Depth (Pixels)
Noise Floor Estimate
500
450
400
350
300
250
200
150
100
50
0
-84 -82 -80 -78 -76 -74 -72 -70 -68
RMS Min

Figure 21

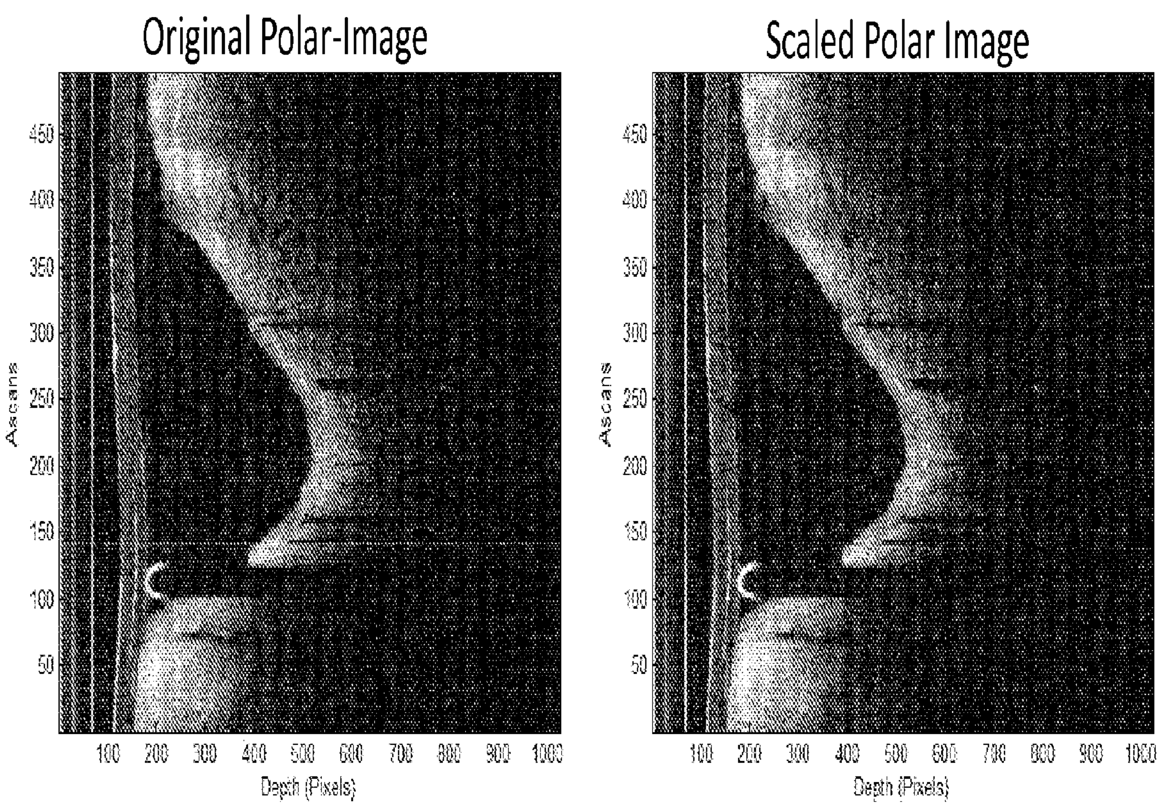
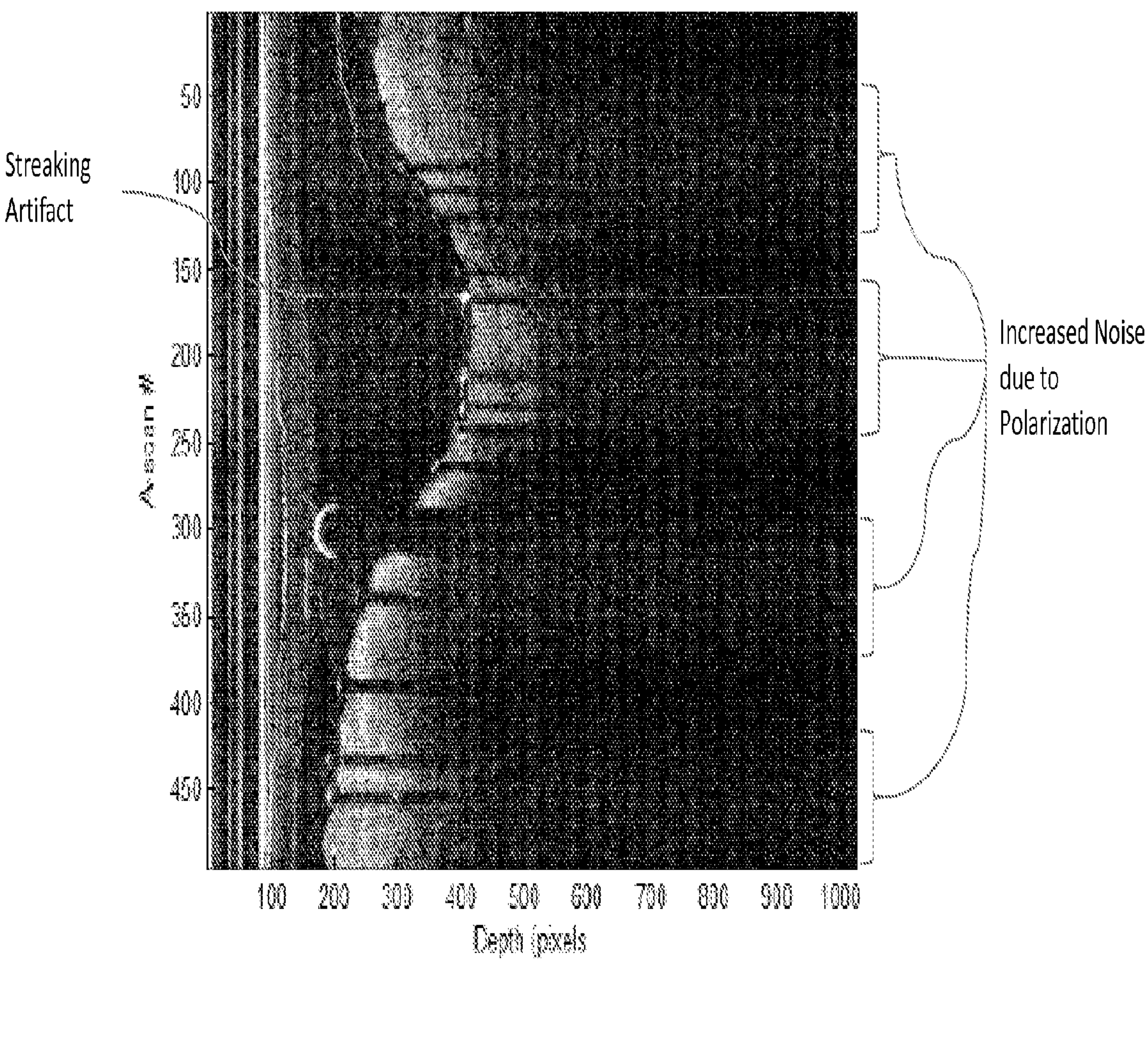

Original Polar-Image
Scaled Polar Image
Ascans
450
400
350
300
250
200
150
100
50
100 200 300 400 500 600 700 800 900 1000
Depth (Pixels)

Figure 22


Streaking
Artifact
Increased Noise
due to
Polarization
50
100
150
200
250
300
350
400
450
A-scan #
100
200
300
400
500
600
700
800
900
1000
Depth (pixels

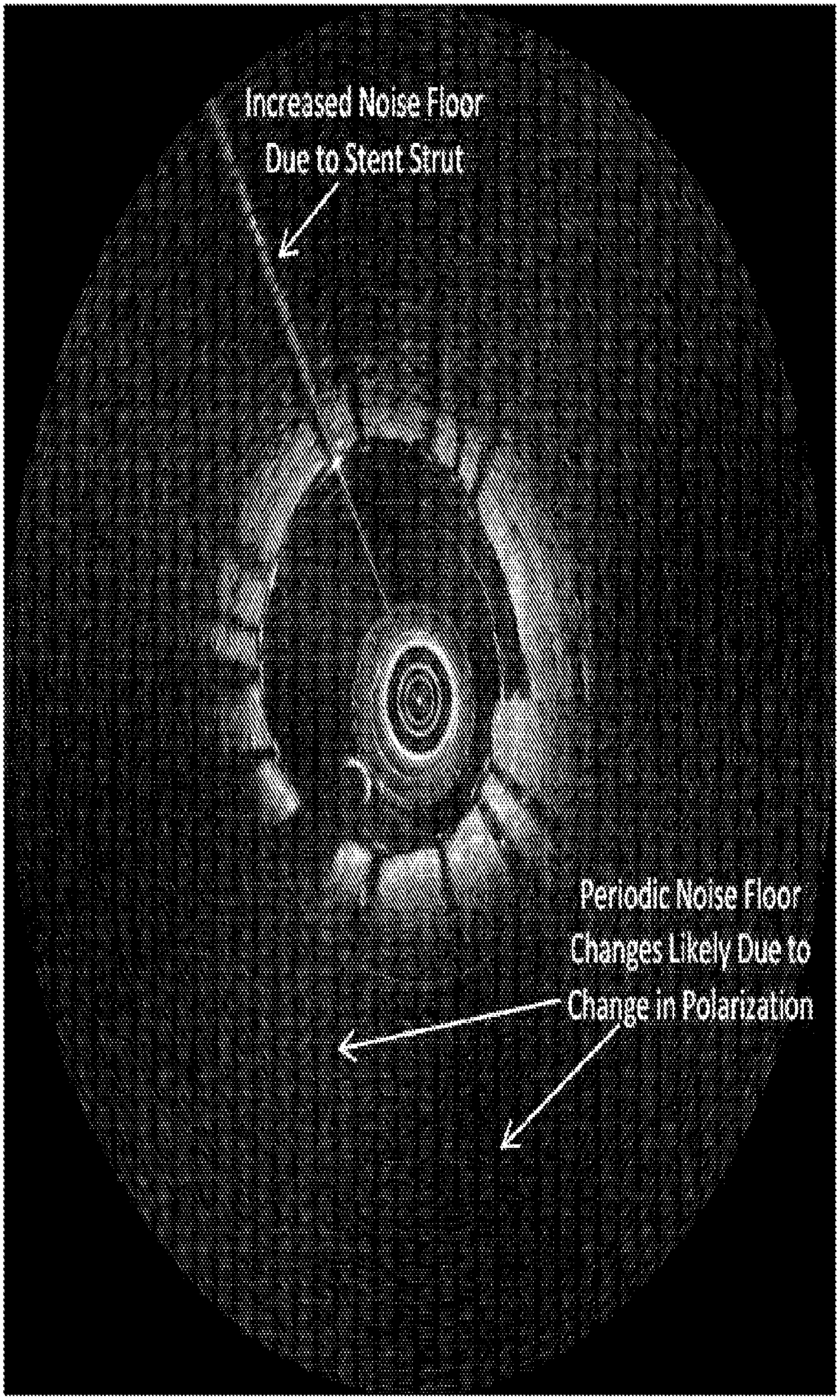

Increased Noise Floor
Due to Stent Strut
Periodic Noise Floor
Changes Likely Due to
Change in Polarization

Figure 23

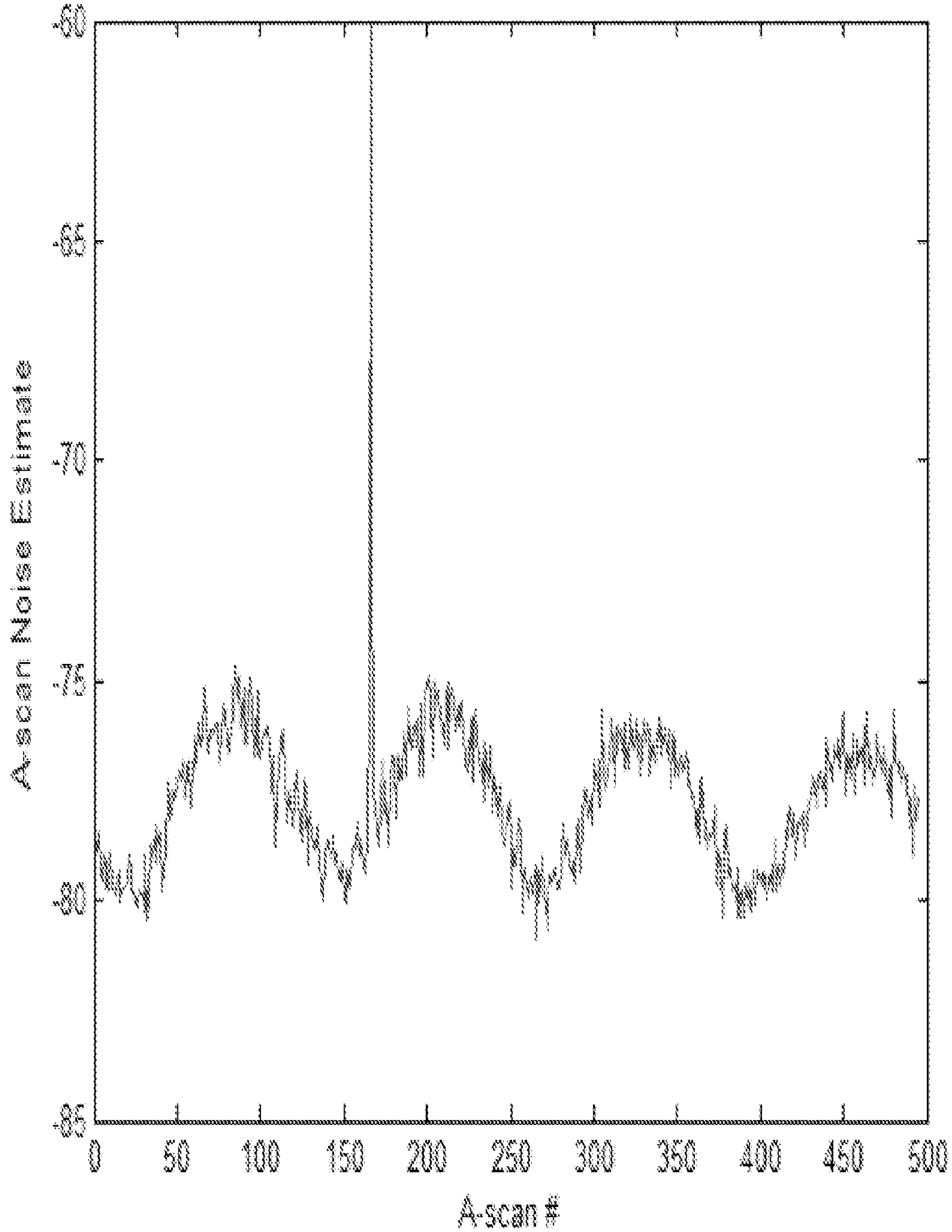

-60
-65
-70
-75
-80
-85
A-scan Noise Estimate
0
50
100
150
200
250
300
350
400
450
500
A-scan #

Figure 24

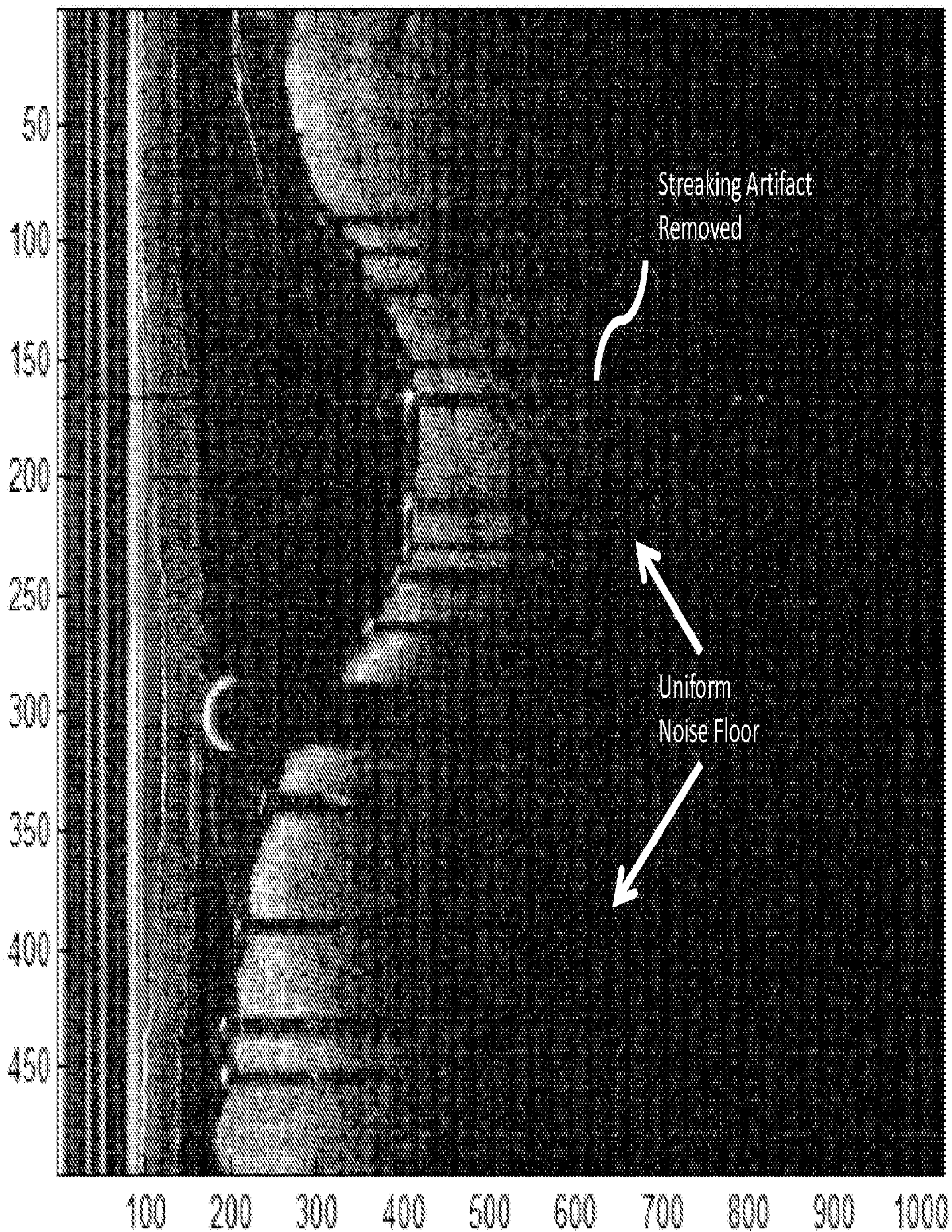

Streaking Artifact
Removed
Uniform
Noise Floor
50
100
150
200
250
300
350
400
450
100
200
300
400
500
600
700
800
900
1000

Figure 25

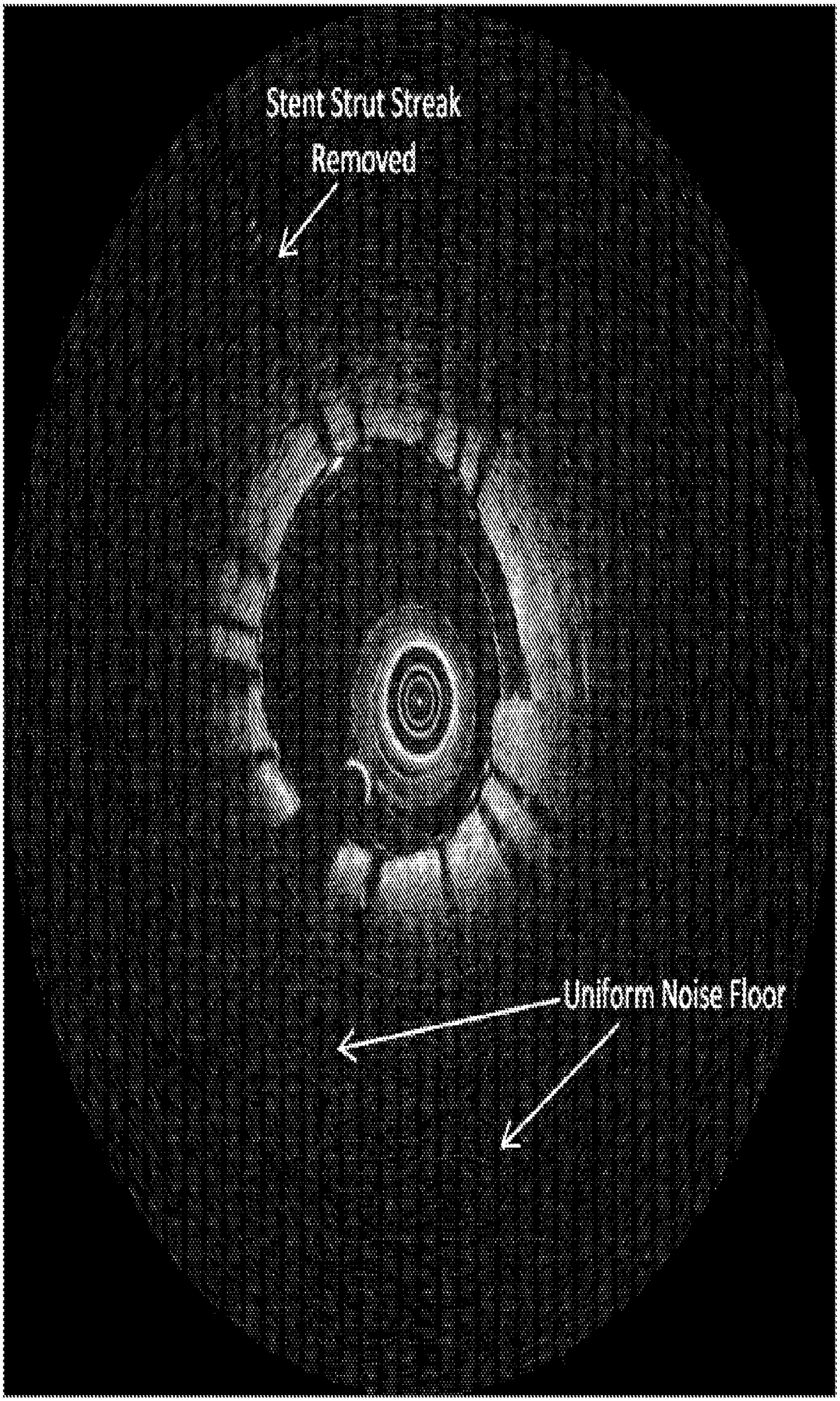
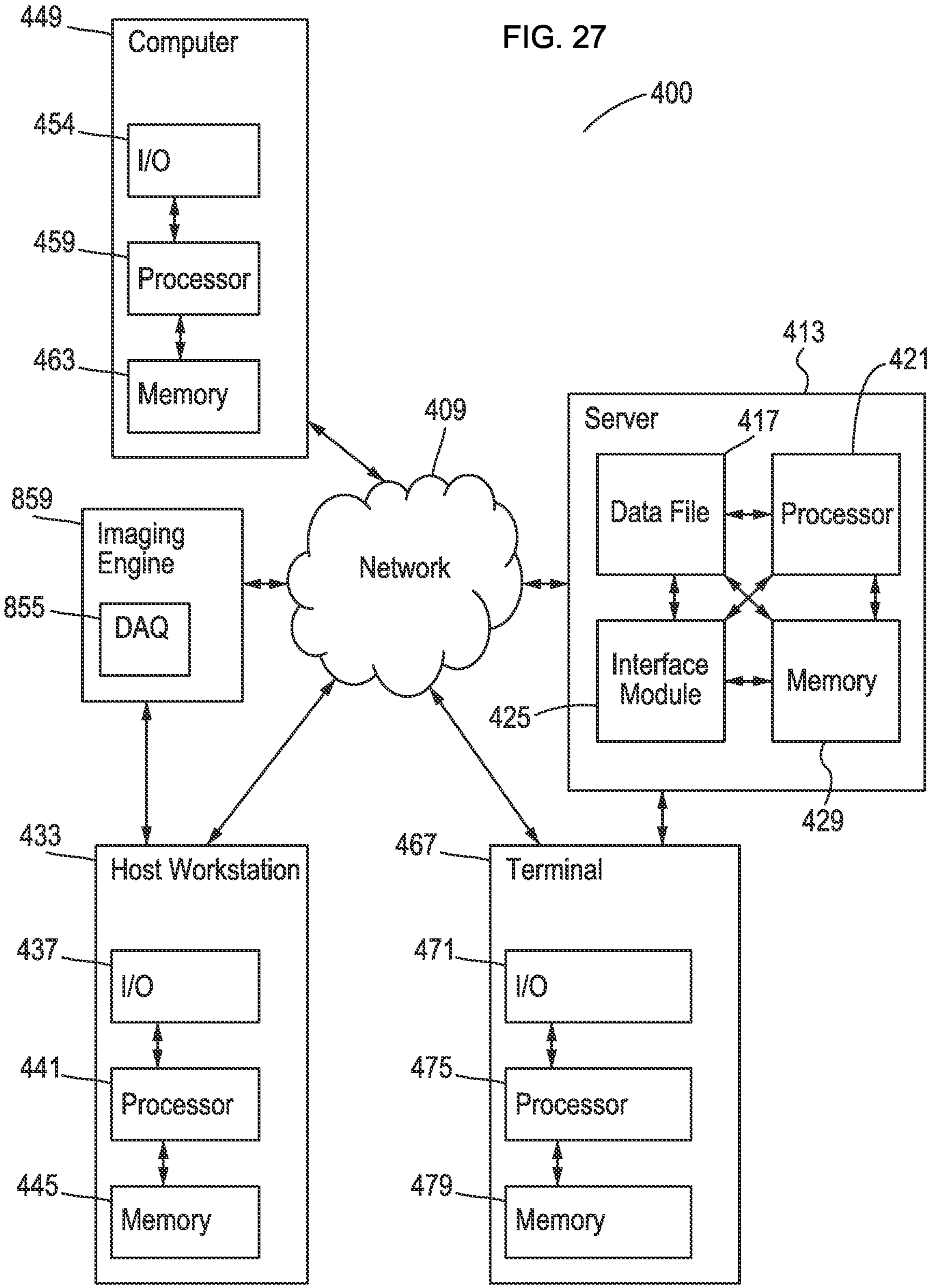


Figure 26

FIG. 27


400
449
Computer
454
I/O
459
Processor
463
Memory
409
Network
413
421
Server
417
Data File
Processor
Interface Module
Memory
425
429
859
Imaging Engine
855
DAQ
433
Host Workstation
437
I/O
441
Processor
445
Memory
467
Terminal
471
I/O
475
Processor
479
Memory

REMOVAL OF A-SCAN STREAKING ARTIFACT

RELATED APPLICATION

The present application claims the benefit of and priority to U.S. Provisional No. 61/710,415, filed Oct. 5, 2012, which is incorporated by reference in its entirety.

TECHNICAL FIELD

This invention generally relates to systems and methods for reducing streaking artifacts and periodic noise in medical imaging.

BACKGROUND

Tomographic imaging is a signal acquisition and processing technology that allows for high-resolution cross-sectional imaging in biological systems. Tomographic imaging systems include, for example, optical coherence tomography systems, ultrasound imaging systems, and computed tomography. Tomographic imaging is particularly well-suited for imaging the subsurface of a vessel or lumen within the body, such as a blood vessel, using probes disposed within a catheter through a minimally invasive procedure.

Typical tomographic imaging catheters consist of an imaging core that rotates and moves longitudinally through a blood vessel, while recording an image video loop of the vessel. The motion results in a 3D dataset, where each frame provides a 360 degree slice of the vessel at different longitudinal section. Each frame, or image, consists of a set of A-lines, which are depth profiles produced by the reflected energy as a function of time.

These frames provide cardiologists with invaluable information such as the location and severity of the stenosis in a patient, the presence of vulnerable plagues, mal-apposed stents, and changes in the disease over time. The information also assists in determining the appropriate treatment plan for the patient, such as drug therapy, stent placement, angioplasty, or bypass surgery. Because a physician is relying on the quality of the image for diagnosis and course of treatment, image quality is critical.

A drawback of tomographic imaging and other signal acquisition imaging technologies is the presence of noise that disrupts the signal and reduces image quality. Noise cause by attenuating objects, such as metal stents, degrades image quality and results in high amplitude signals that appear as an actual streak within the obtained images. The resulting streak is often called a streaking artifact. In addition, periodic noise, such as noise caused by changes in polarization, can also degrade image quality. The reduced image quality caused by both streaking artifacts and periodic noise impedes a physician's ability to accurately interpret the medical image.

Typically, noise is reduced in medical images using filtering and averaging techniques. Filtering techniques include applying digital filters, such as mean and median, wavelet, anisotropic and bi-lateral filters, uniformly across A-scans. Typical averaging techniques treat noise as a uniform background disruption across an imaging data set and average the noise across A-scans. Although filters and uniform signal averaging successfully remove some noise, such techniques are inefficient at removing streaking artifacts and periodic noise.

Current techniques aimed at removing streaking artifacts are complicated and inefficient due to multiple complicated processing techniques. One technique to remove streaking artifacts utilizes nearest-neighbor pattern recognition. Using nearest neighbor pattern recognition, A-scans that intersect an attenuating object responsible for the streaking are detected. After the A-scans are detected, the A-scans are removed from the data set and the streaking artifact is replaced with interpolated data. Another technique requires obtaining repeated B-scans of a target location, registering and aligning the obtained B-scans, and averaging the A-scans across the B-scans using a weighted average. This technique is only successful in removing streaking artifacts if the attenuating object does not produce a streaking artifact uniformly across the B-scans.

SUMMARY

Systems and methods of the invention provide for a fast and simple processing technique for reducing noise, such as streaking artifacts, in tomographic images. Unlike other averaging techniques that estimate noise across A-scans, noise is treated as a gain that is specific and unique to each A-scan. By treating noise as a gain that is specific and unique to each A-scan and independent from other A-scans in an imaging data set, an algorithm estimates and scales the noise within each individual A-scan. The combination of the individually scaled A-scans results in 2-D and 3-D images without streaking artifacts and periodic noise. This advantageously and efficiently overcomes the reduced image quality associated with streaking artifacts without the need for repeated imaging of a target location and multiple, complicated processing techniques. Moreover, the simple noise reduction technique improves the ability for a physician to accurately interpret a medical image.

Tomographic imaging systems suitable for use in the invention include, for example, ultrasound imaging systems, optical coherence tomography systems, and combined OCT-ultrasound imaging systems.

In one aspect, noise is reduced in an image by obtaining an A-scan from a data imaging set. The A-scan has a signal and the signal defines an amplitude. The amplitude of an A-scan is a depth profile of the reflected signal. In OCT systems, an A-scan is a depth profile produced by the reflected optical energy as a function of time. To remove noise from an A-scan, a noise floor specific to the A-scan is estimated and the amplitude of the signal is scaled based on the estimated noise floor.

In one embodiment, the noise floor of an A-scan is estimated by averaging the amplitude of the signal at a plurality of pixels within a window of the A-scan image. The A-scan is then normalized to reduce noise by scaling the amplitude of the A-scan based on the A-scan's specific estimated noise floor. Suitable averaging techniques for estimating the noise floor include, but are not limited to, computing the median, mode, arithmetic mean, geometric mean, harmonic mean, quadratic mean, also known as root mean square, and any weighted means of the A-scan signal within a window.

In another embodiment, the estimated noise floor is determined by performing a moving average across an A-scan signal. The moving average is computed across the A-scan using a sliding window. The sliding window can vary in pixel width. For example, the sliding window can be a 10 pixel sliding window or a 100 pixel sliding window. Suitable averaging techniques for performing the moving average include, but are not limited to, computing the median, mode, arithmetic mean, geometric mean, harmonic mean, quadratic mean, also known as root mean square, and any weighted means of each data set within the sliding window. The minimum average value of the A-scan signal determined from the moving average is used as the estimated noise floor for the

A-scan. The A-scan is then normalized to reduce noise by scaling the amplitude based on the estimated noise floor.

To create a B-scan or scan-converted image with reduced noise, an estimated noise floor is individually estimated for each A-scan of the B-scan and scaled based on its specific estimated noise floor. For example, in one aspect, a plurality of A-scans is obtained from an imaging data set. Each A-scan of the plurality of A-scans has a signal and the signal defines an amplitude. In each A-scan, an elevated noise floor specific to the A-scan is estimated. In certain aspects, the estimated noise floor is the average amplitude of the A-scan. In another aspect, the noise floor is estimated by performing a moving average across the A-scan, and the minimum average amplitude of the A-scan computed from the moving average is the estimated noise floor. To eliminate the noise within each A-scan, the amplitude of the signal within each A-scan image is scaled by its specific estimated noise floor. The scaled A-scans are then combined to form a B-scan image having reduced noise. The B-scan image is scan-converted into a Cartesian coordinate system to create a final tomographic view of the image with reduced noise.

Other and further aspects and features of the invention will be evident from the following detailed description and the accompanying drawings, which are intended to illustrate, not limit, the invention.

BRIEF DESCRIPTION OF DRAWINGS

FIG. **1** is a perspective view of a vessel.

FIG. **2** is a cross sectional view of the vessel shown in FIG. **1**.

FIG. **3** is a diagram of components of an optical coherence tomography (OCT) system.

FIG. **4** is a diagram of the imaging engine shown in FIG. **3**.

FIG. **5** is a diagram of a light path in an OCT system of certain embodiments of the invention.

FIG. **6** is a patient interface module of an OCT system.

FIG. **7** is an illustration of the motion of parts of an imaging catheter according to certain embodiments of the invention.

FIG. **8** shows an array of A scan lines of a three-dimensional imaging system according to certain embodiments of the invention.

FIG. **9** shows the positioning of A scans with in a vessel.

FIG. **10** illustrates a set of A scans used to compose a B scan according to certain embodiments of the invention.

FIG. **11** shows the set of A scans shown in FIG. **10** within a cross section of a vessel.

FIG. **12** shows a sample OCT B-Scan polar image calculated from 660 A-scans.

FIG. **13** shows a scan-converted OCT image from the B-scan of FIG. **12**.

FIG. **14** depicts a block diagram for reducing noise according to methods of the invention.

FIG. **15** shows an original polar image of a vessel with a streaking artifact.

FIG. **16** shows A-scan **350** of the polar image of FIG. **15**.

FIG. **17** depicts the squared amplitude of the signal in A-scan **350**.

FIG. **18** depicts the mean squared amplitude of A-Scan **350** across 100 pixel sliding window.

FIG. **19** shows the calculated root mean square of the A-scan **350**.

FIG. **20** shows the comparison between the original polar image of the B-scan and the noise floor estimate individually computed for each A-scan.

FIG. **21** shows the original polar image in comparison to the scaled polar image having the streaking artifact removed.

FIG. **22** shows a polar image having periodic noise across the A-scans.

FIG. **23** shows the scan-converted image of FIG. **22**.

FIG. **24** plots the estimated noise floor estimate for each A-scan from the polar image of FIG. **22**.

FIG. **25** shows the polar image of FIG. **22** scaled by the estimated noise floors of FIG. **24**.

FIG. **26** shows the scan-converted image of FIG. **25**.

FIG. **27** is a system diagram according to certain embodiments.

DESCRIPTION

This invention generally relates to reducing noise in tomographic medical images. Medical tomographic imaging is a general technology class in which sectional and multidimensional anatomic images are constructed from acquired data. The data can be collected from a variety of signal acquisition systems including, but not limited to, magnetic resonance imaging (MRI), radiography methods including fluoroscopy, x-ray tomography, computed axial tomography and computed tomography, nuclear medicine techniques such as scintigraphy, positron emission tomography and single photon emission computed tomography, photo acoustic imaging ultrasound devices and methods including, but not limited to, intravascular ultrasound spectroscopy (IVUS), ultrasound modulated optical tomography, ultrasound transmission tomography, other tomographic techniques such as electrical capacitance, magnetic induction, functional MRI, optical projection and thermo-acoustic imaging, combinations thereof and combinations with other medical techniques that produce one-, two- and three-dimensional images. Although the exemplifications described herein are drawn to the invention as applied to OCT, at least all of these techniques are contemplated for use with the systems and methods of the present invention.

Systems and methods of the invention have application in intravascular imaging methodologies such as intravascular ultrasound (IVUS) and optical coherence tomography (OCT) among others that produce a three-dimensional image of a lumen. A segment of a lumen **101** is shown in FIG. **1** having a feature **113** of interest. FIG. **2** shows a cross-section of lumen **101** through feature **113**. In certain embodiments, intravascular imaging involves positioning an imaging device near feature **113** and collecting data representing a three-dimensional image.

OCT is a medical imaging methodology using a specially designed catheter with a miniaturized near infrared light-emitting probe attached to the distal end of the catheter. As an optical signal acquisition and processing method, it captures micrometer-resolution, three-dimensional images from within optical scattering media (e.g., biological tissue). Commercially available OCT systems are employed in diverse applications, including art conservation and diagnostic medicine, notably in ophthalmology where it can be used to obtain detailed images from within the retina. The detailed images of the retina allow one to identify several eye diseases and eye trauma. Recently it has also begun to be used in interventional cardiology to help diagnose coronary artery disease. OCT allows the application of interferometric technology to see from inside, for example, blood vessels, visualizing the endothelium (inner wall) of blood vessels in living individuals.

Other applications of OCT and other signal processing imaging systems for biomedical imaging include use in: dermatology in order to image subsurface structural and blood flow formation; dentistry in order to image the structure of teeth and gum line to identify and track de-mineralization and re-mineralization, tarter, caries, and periodontal disease; gastroenterology in order to image the gastrointestinal tract to detect polyps and inflammation, such as that caused by Crohn's disease and ulcerative colitis; cancer diagnostics in order to discriminate between malignant and normal tissue.

Generally, an OCT system comprises three components which are 1) an imaging catheter 2) OCT imaging hardware, 3) host application software. When utilized, the components are capable of obtaining OCT data, processing OCT data, and transmitting captured data to a host system. OCT systems and methods are generally described in Castella et al., U.S. Pat. No. 8,108,030, Milner et al., U.S. Patent Application Publication No. 2011/0152771, Condit et al., U.S. Patent Application Publication No. 2010/0220334, Castella et al., U.S. Patent Application Publication No. 2009/0043191, Milner et al., U.S. Patent Application Publication No. 2008/0291463, and Kemp, N., U.S. Patent Application Publication No. 2008/0180683, the content of each of which is incorporated by reference in its entirety. In certain embodiments, systems and methods of the invention include processing hardware configured to interact with more than one different three dimensional imaging system so that the tissue imaging devices and methods described here in can be alternatively used with OCT, IVUS, or other hardware.

Various lumen of biological structures may be imaged with aforementioned imaging technologies in addition to blood vessels, including, but not limited, to vasculature of the lymphatic and nervous systems, various structures of the gastrointestinal tract including lumen of the small intestine, large intestine, stomach, esophagus, colon, pancreatic duct, bile duct, hepatic duct, lumen of the reproductive tract including the vas deferens, vagina, uterus and fallopian tubes, structures of the urinary tract including urinary collecting ducts, renal tubules, ureter, and bladder, and structures of the head and neck and pulmonary system including sinuses, parotid, trachea, bronchi, and lungs.

The arteries of the heart are particularly useful to examine with imaging devices such as OCT. OCT imaging of the coronary arteries can determine the amount of plaque built up at any particular point in the coronary artery. The accumulation of plaque within the artery wall over decades is the setup for vulnerable plaque which, in turn, leads to heart attack and stenosis (narrowing) of the artery. OCT is useful in determining both plaque volume within the wall of the artery and/or the degree of stenosis of the artery lumen. It can be especially useful in situations in which angiographic imaging is considered unreliable, such as for the lumen of ostial lesions or where angiographic images do not visualize lumen segments adequately. Example regions include those with multiple overlapping arterial segments. It is also used to assess the effects of treatments of stenosis such as with hydraulic angioplasty expansion of the artery, with or without stents, and the results of medical therapy over time. In an exemplary embodiment, the invention provides a system for capturing a three dimensional image by OCT.

In OCT, a light source delivers a beam of light to an imaging device to image target tissue. Light sources can include pulsating light sources or lasers, continuous wave light sources or lasers, tunable lasers, broadband light source, or multiple tunable laser. Within the light source is an optical amplifier and a tunable filter that allows a user to select a wavelength of light to be amplified. Wavelengths commonly used in medical applications include near-infrared light, for example between about 800 nm and about 1700 nm.

Methods of the invention apply to image data obtained from obtained from any OCT system, including OCT systems that operate in either the time domain or frequency (high definition) domain. Basic differences between time-domain OCT and frequency-domain OCT is that in time-domain OCT, the scanning mechanism is a movable minor, which is scanned as a function of time during the image acquisition. However, in the frequency-domain OCT, there are no moving parts and the image is scanned as a function of frequency or wavelength.

In time-domain OCT systems an interference spectrum is obtained by moving the scanning mechanism, such as a reference minor, longitudinally to change the reference path and match multiple optical paths due to reflections within the sample. The signal giving the reflectivity is sampled over time, and light traveling at a specific distance creates interference in the detector. Moving the scanning mechanism laterally (or rotationally) across the sample produces two-dimensional and three-dimensional images.

In frequency domain OCT, a light source capable of emitting a range of optical frequencies excites an interferometer, the interferometer combines the light returned from a sample with a reference beam of light from the same source, and the intensity of the combined light is recorded as a function of optical frequency to form an interference spectrum. A Fourier transform of the interference spectrum provides the reflectance distribution along the depth within the sample.

Several methods of frequency domain OCT are described in the literature. In spectral-domain OCT (SD-OCT), also sometimes called "Spectral Radar" (Optics letters, Vol. 21, No. 14 (1996) 1087-1089), a grating or prism or other means is used to disperse the output of the interferometer into its optical frequency components. The intensities of these separated components are measured using an array of optical detectors, each detector receiving an optical frequency or a fractional range of optical frequencies. The set of measurements from these optical detectors forms an interference spectrum (Smith, L. M. and C. C. Dobson, Applied Optics 28: 3339-3342), wherein the distance to a scatterer is determined by the wavelength dependent fringe spacing within the power spectrum. SD-OCT has enabled the determination of distance and scattering intensity of multiple scatters lying along the illumination axis by analyzing a single the exposure of an array of optical detectors so that no scanning in depth is necessary. Typically the light source emits a broad range of optical frequencies simultaneously. Alternatively, in swept-source OCT, the interference spectrum is recorded by using a source with adjustable optical frequency, with the optical frequency of the source swept through a range of optical frequencies, and recording the interfered light intensity as a function of time during the sweep. An example of swept-source OCT is described in U.S. Pat. No. 5,321,501.

Generally, time domain systems and frequency domain systems can further vary in type based upon the optical layout of the systems: common beam path systems and differential beam path systems. A common beam path system sends all produced light through a single optical fiber to generate a reference signal and a sample signal whereas a differential beam path system splits the produced light such that a portion of the light is directed to the sample and the other portion is directed to a reference surface. Common beam path systems are described in U.S. Pat. Nos. 7,999,938; 7,995,210; and 7,787,127 and differential beam path systems are described in U.S. Pat. Nos. 7,783,337; 6,134,003; and 6,421,164, the contents of each of which are incorporated by reference herein in its entirety.

In certain embodiments, the invention provides a differential beam path OCT system with intravascular imaging capability as illustrated in FIG. **3**. For intravascular imaging, a light beam is delivered to the vessel lumen via a fiber-optic based imaging catheter **826**. The imaging catheter is connected through hardware to software on a host workstation. The hardware includes an imagining engine **859** and a handheld patient interface module (PIM) **839** that includes user controls. The proximal end of the imaging catheter is connected to PIM **839**, which is connected to an imaging engine as shown in FIG. **3**. As shown in FIG. **4**, the imaging engine **859** (e.g., a bedside unit) houses a power supply **849**, light source **827**, interferometer **831**, and variable delay line **835** as well as a data acquisition (DAQ) board **855** and optical controller board (OCB) **851**. A PIM cable **841** connects the imagine engine **859** to the PIM **839** and an engine cable **845** connects the imaging engine **859** to the host workstation.

FIG. **5** shows light path in a differential beam path system according to an exemplary embodiment of the invention. Light for image capture originates within the light source **827**. This light is split between an OCT interferometer **905** and an auxiliary, or "clock", interferometer **911**. Light directed to the OCT interferometer is further split by splitter **917** and recombined by splitter **919** with an asymmetric split ratio. The majority of the light is guided into the sample path **913** and the remainder into a reference path **915**. The sample path includes optical fibers running through the PIM **839** and the imaging catheter **826** and terminating at the distal end of the imaging catheter where the image is captured.

Typical intravascular OCT involves introducing the imaging catheter into a patient's target vessel using standard interventional techniques and tools such as a guide wire, guide catheter, and angiography system. The imaging catheter may be integrated with IVUS by an OCT-IVUS system for concurrent imaging, as described in, for example, Castella et al. U.S. Patent Application Publication No. 2009/0043191 and Dick et al. U.S. Patent Application Publication No. 2009/0018393, both incorporated by reference in their entirety herein.

Rotation of the imaging catheter is driven by spin motor **861** while translation is driven by pullback motor **865**, shown in FIG. **6**. This results in a motion for image capture described by FIG. **7**. Blood in the vessel is temporarily flushed with a clear solution for imaging. When operation is triggered from the PIM or control console, the imaging core of the catheter rotates while collecting image data. Using light provided by the imaging engine, the inner core sends light into the tissue in an array of A-scan lines as illustrated in FIG. **8** and detects reflected light.

FIG. **9** shows the positioning of A-scans within a vessel. Each place where one of A-scans A**11**, A**12**, . . . , AN intersects a surface of a feature within vessel **101** (e.g., a vessel wall) coherent light is reflected and detected. Catheter **826** translates along axis **117** being pushed or pulled by pullback motor **865**.

The reflected, detected light is transmitted along sample path **913** to be recombined with the light from reference path **915** at splitter **919** (FIG. **5**). A variable delay line (VDL) **925** on the reference path uses an adjustable fiber coil to match the length of reference path **915** to the length of sample path **913**. The reference path length is adjusted by a stepper motor translating a minor on a translation stage under the control of firmware or software. The free-space optical beam on the inside of the VDL **925** experiences more delay as the minor moves away from the fixed input/output fiber.

The combined light from splitter **919** is split into orthogonal polarization states, resulting in RF-band polarization-diverse temporal interference fringe signals. The interference fringe signals are converted to photocurrents using PIN photodiodes **929***a*, **929***b*, . . . on the OCB **851** as shown in FIG. **5**. The interfering, polarization splitting, and detection steps are done by a polarization diversity module (PDM) on the OCB. Signal from the OCB is sent to the DAQ **855**, shown in FIG. **4**. The DAQ includes a digital signal processing (DSP) microprocessor and a field programmable gate array (FPGA) to digitize signals and communicate with the host workstation and the PIM. The FPGA converts raw optical interference signals into meaningful OCT images. The DAQ also compresses data as necessary to reduce image transfer bandwidth to 1 Gbps (e.g., compressing frames with a glossy compression JPEG encoder).

Data is collected from A-scans A**11**, A**12**, . . . , AN and stored in a tangible, non-transitory memory. Typically, rotational systems consist of an imaging core which rotates and pulls back (or pushes forward) while recording an image video loop. This motion results in a three dimensional dataset of two dimensional image frames, where each frame provides a 360° slice of the vessel at different longitudinal locations.

A set of A-scans captured in a helical pattern during one rotation of catheter **826** around axis **117** collectively define a B scan. FIG. **10** illustrates a set of A-scans A**11**, A**12**, . . . , A**18** used to form a B scan according to certain embodiments of the invention. These A-scan lines are shown as would be seen looking down axis **117** (i.e., longitudinal distance between then is not shown). While eight A-scan lines are illustrated in FIG. **10**, typical OCT applications can include between 300 and 1,000 A-scan lines to create a B scan (e.g., about 660). Reflections detected along each A-scan line are associated with features within the imaged tissue. Reflected light from each A-scan is combined with corresponding light that was split and sent through reference path **915** and VDL **925** and interference between these two light paths as they are recombined indicates features in the tissue.

The data of all the A-scan lines together represent a three-dimensional image of the tissue. The data of the A-scan lines generally referred to as a B scan can be used to create an image of a cross section of the tissue, sometimes referred to as a tomographic view. For example, FIG. **11** shows the set of A-scans shown in FIG. **10** within a cross section of a vessel. The set of A-scans obtained by rotational imaging modality can be combined to form a B-scan. FIG. **12** is an example of an OCT polar coordinate B-Scan with 660 A-scans. To create a final tomographic view of the vessel, the B-scan is scan converted to a Cartesian coordinate system. FIG. **13** displays the scan converted image of the B-scan in FIG. **12**.

OCT imaging and other signal processing techniques are often affected by noise that disrupts the A-scan signals. Noise left unprocessed can easily distort the resulting B-scan and scan-converted OCT images. Noise can be attributed to streaking artifacts. Streaking artifacts are inadvertent reflections in the path of the light source caused by attenuating objects, such as those due to a stent strut. Other sources of noise include periodic noise due to changes in polarization, interference between coherent waves backscattered from nearby scatters in a sample, and unknown factors. In the case of spectra-domain OCT, the readout electronics of the array detector often adds periodic noise to the recorded optical intensities. In swept-source OCT, electronic clocks and counters can produce a similar periodic noise.

Systems and methods of the invention include image-processing techniques to reduce noise and remove streaking artifacts within medical images by scaling the noise within one-dimensional images, or depth resolved A-scans, of an imaging data set. By reducing the noise individually at the A-scan level, noise is effectively reduced in the B-scan and scan-converted image. Although the following description is directed towards reducing noise in OCT images, methods and systems of invention can be utilized to reduce noise using one-dimensional images obtained from any medical signal-processing imaging technique.

Described below are methods of the invention for reducing noise in an A-scan following the block diagram of FIG. **14**. It should be noted that the following embodiments can be manually computed to or automatically computed and executed by a processor of a computing system to reduce noise. Automatic computation reduces error associated with manual computation.

FIG. **14** depicts a block diagram for reducing noise according to methods of the invention. Step **300** involves obtaining an A-scan from an imaging data set. Step **302** involves estimating a noise floor specific to the A-scan. Step **306** involves scaling, or normalizing, the noise within the A-scan based on the specific estimated noise floor.

The estimated noise floor, an intensity value in dB, is treated as a gain that is constant across the A-scan. Therefore, in order to remove the effects of this gain, the noise floor estimate is subtracted from the original image. This results in an A-scan with reduced noise.

Methods for obtaining a depth resolved A-scan are well known in the art and vary depending on the signal processing imaging technique used. The A-scan is a depth resolved image of the interference signal. For time-domain OCT systems, the depth resolved A-scan can be obtained by taking the square root of the reflectivity of the interference signal versus depth. In certain aspects, the basic operation to achieve a depth resolved A-scan from the interference fringe signal is performing a fast Fourier transform on the imaging data. For example, typically in Fourier Domain OCT, including spectral domain OCT, the interference signal is sent to an optical spectrometer and detected as a function of optical frequency. With a fixed optical delay in the reference arm, light reflected from different sample depths produces interference patterns with the different frequency components. A Fourier transform is then used to resolve different depth reflections, thereby generating a depth profile of the sample (A-scan).

The reflected energy in a depth resolved A-scan appears as a signal, or a line, across the x-axis within the A-scan. The signal is represented by a row of pixels along the x-axis of the A-scan. The y-coordinate of each pixel is the amplitude of the signal at that pixel and the x-coordinate is the index for pixel depth.

After an A scan is obtained from an imaging data set, Step **302** requires estimating the noise floor of the A-scan. In certain aspects, estimating the noise floor involves averaging the amplitude signal across the A-scan. For example, noise is determined by averaging the amplitude of the pixels across a window of size N within the A-scan. N is the number of pixels across the window. Suitable averaging techniques for estimating the noise floor include, but are not limited to, computing the median, mode, arithmetic mean, geometric mean, harmonic mean, quadratic mean, also known as root mean square, and any weighted means of the A-scan signal within a window.

Any window of size N can be used to average noise within the A-scan. For example, the window can span across all of the pixels within an A-scan, or can include a window of 10 pixels or a window of 100 pixels. The number of pixels chosen for the window size determines the range and extent noise is reduced.

In some embodiments, the calculated average noise within the window of size N is the estimated noise floor. The A-scan is then scaled by the A-scan's specific estimated noise floor, as in step **306**. The calculated average is treated a gain that is constant across the A-scan and is subtracted from the amplitude of the original A-scan. In other words, the amplitude at each pixel is reduced by the calculated average.

In certain aspects, the A-scan image is linearized prior to averaging. Once the linearized signal is averaged, the calculated average is converted back to the logarithmic scale. The converted average is the estimated noise floor.

According to aspects of the invention, estimating the noise floor for step **304** includes computing a moving average across an A-scan signal. The moving average involves calculating averaging pixels within a siding window of size N across an A-scan. N is the pixel width of the sliding window. The pixel width N can be any number of pixels, for example a 10 pixel sliding window, 50 pixel sliding window, or a 100 window sliding window. Suitable averaging techniques for performing the moving average include, but are not limited to, computing the median, mode, arithmetic mean, geometric mean, harmonic mean, quadratic mean, also known as root mean square, and any weighted means of each pixel data set within the sliding window.

After performing the moving average across the A-scan, the resulting averages are used as the basis for the estimated noise floor. In certain aspects, the minimum average is used as the scale factor. In other aspects, a mean, median, or mode of the resulting averages is utilized as the scale factor. Once the estimated noise floor is obtained, the amplitude of the A-scan is scaled by the reducing the amplitude by the estimated noise floor, as in step **306**. If the A-scan data was linearized, the estimated noise floor is converted back to the logarithmic scale prior to scaling the A-scan amplitude.

In certain aspects, a plurality of A-scans is obtained from an imaging data set. The steps outlined in FIG. **14** and described above, are then repeated for each A-scan of the plurality of A-scans. As a result, noise is specific to each A-scan of the plurality of A-scan is estimated and each A-scan is scaled based on its specific estimated noise floor. The A-scans with reduced noise are combined to create a B-scan with reduced noise. To create the final tomographic view with reduced noise, the B-scan is scan-converted to the Cartesian coordinate system.

In some embodiments, a constant noise floor is added back into each A-scan of a plurality of A-scans. The constant noise floor is the same across all A-scans. The constant noise floor can include, for example, the average amplitude across the plurality of A-scans.

The following example shows a method of practicing the invention.

FIG. **15** depicts an original polar image, or B-scan. In FIG. **15**, noise the form of a streaking artifact is displayed in the A-scans near A-scan **150**. In order to remove the effects of this noise and any other noise within the B-scan, an estimated noise floor for each A-scan is calculated. The following figures and steps show the estimated noise floor calculation for A-scan **350** using a moving root mean square average across A-scan **405** with a sliding window of pixel size 100.

FIG. **16** depicts A-scan **350** of the original polar image of FIG. **15**. The first step is to square the amplitude of each pixel within the A-scan **350** using equation 1, where $Ascan_{dB}$ is the amplitude of the signal in the A-scan in dB. Although not exemplified, prior to squaring the amplitude and performing the moving average, the A-scan can be linearized using Equation 2.

$$Ascan_{sqr} = Ascan_{dB}^2 \qquad \text{Equation 1}$$

-continued

$$Ascan_{lin} = 1^{-\left(\frac{Ascan_{dB}}{10}\right)} \qquad \text{Equation 1}$$

FIG. **17** depicts the squared amplitude of the signal in A-scan **350**. Next, the mean square of the amplitude of A-scan **350** is determined using a sliding window of 100 pixel using equations 3 and 4, where N is size of the sliding window and MS is the mean squared.

$$F(1:N) = \frac{1}{N} \qquad \text{Equation 3}$$

$$MS = Ascan_{sqr} * F \qquad \text{Equation 4}$$

FIG. **18** depicts the calculated mean square of the A-scan **350** using the 100 pixel sliding window. After the mean square is calculated across the A-scan **350**, the root mean square of the amplitude is computed as in equation 5.

$$RMS = \sqrt{MS} \qquad \text{Equation 5}$$

FIG. **19** shows the calculated root mean square of the A-scan **350**. As shown in FIG. **19**, the minimum value of the root mean square for A-scan **350** equals −80.5 dB. This minimum average value is the estimated noise floor for the A-scan **350**.

The above calculations for estimating a noise floor for A-scan **350** are then repeated for each A-scan in the data set. FIG. **20** shows the comparison between the original polar image and the noise floor estimates individually computed for each A-scan. As shown in the comparison, the estimated noise floor is highest at the A-scans near A-scan **150** that exhibit a streaking artifact.

FIG. **21** shows the original polar image in comparison to the scaled polar image with the streaking artifact removed. In order to form the scaled polar image, each A-scan is scaled by its estimated noise floor using equation 6, where the estimated noise floor of an A-scan is S_{ascan}.

$$Ascan_{scaled} = Ascan_{dB} - S_{ascan} \qquad \text{Equation 6}$$

The scaled polar image shown in FIG. **21** is obtained by combining each of the individually scaled A-scan.

If the A-scan data was linearized using equation 2 prior to performing the calculations for equations 1, 3, 4, and 5, the minimum average value of the linearized root mean square should be converted to the logarithmic scale to obtain the estimated noise floor of the A-scan, using equation 7.

$$S_{ascan} = \min(10 \log_{10}(RMS)) \qquad \text{Equation 7}$$

In addition to removing streaking artifacts, methods of the invention also have the effect of reducing periodic noise, such as changes in polarization as the imaging catheter rotates. FIG. **22** depicts a polar image, or B-scan, with periodic noise across the A-scans of the B-scan. FIG. **23** shows the scan-converted image of FIG. **22**. The highlighted regions show higher intensity pixels, which is noise. In addition to periodic noise, FIG. **22** also shows a streaking artifact. Noise was removed from FIG. **22** by following the steps outlined in FIG. **14**. First, an estimated noise floor was calculated for each A-scan in the B-scan. FIG. **24** plots the estimated noise floor specific to each A-scan from the B-scan of FIG. **22**. The high amplitude spike around **170** represents the estimated noise caused by the streaking artifact and the sinusoidal pattern across the A-scans is caused by the periodic noise. Second, the amplitude of each A-scan of the plurality of A-scans is scaled the using the A-scan's specific estimated noise floor to remove the noise. The scaled A-scans are combined to form a B-scan. The resulting normalized B-scan without the streaking artifact and having a uniform noise floor is shown in FIG. **25**. FIG. **26** shows the scan-converted image of FIG. **25**.

In some embodiments, a device of the invention includes an OCT imaging system and obtains a three-dimensional data set through the operation of OCT imaging hardware. In some embodiments, a device of the invention is a computer device such as a laptop, desktop, or tablet computer, and obtains a three-dimensional data set by retrieving it from a tangible storage medium, such as a disk drive on a server using a network or as an email attachment.

Methods of the invention can be performed using software, hardware, firmware, hardwiring, or combinations of any of these. Features implementing functions can also be physically located at various positions, including being distributed such that portions of functions are implemented at different physical locations (e.g., imaging apparatus in one room and host workstation in another, or in separate buildings, for example, with wireless or wired connections).

In some embodiments, a user interacts with a visual interface to view images from the imaging system. Input from a user (e.g., parameters or a selection) are received by a processor in an electronic device. The selection can be rendered into a visible display. An exemplary system including an electronic device is illustrated in FIG. **27**. As shown in FIG. **27**, imaging engine **859** communicates with host workstation **433** as well as optionally server **413** over network **409**. In some embodiments, an operator uses computer **449** or terminal **467** to control system **400** or to receive images. An image may be displayed using an I/O **454**, **437**, or **471**, which may include a monitor. Any I/O may include a keyboard, mouse or touchscreen to communicate with any of processor **421**, **459**, **441**, or **475**, for example, to cause data to be stored in any tangible, nontransitory memory **463**, **445**, **479**, or **429**. Server **413** generally includes an interface module **425** to effectuate communication over network **409** or write data to data file **417**.

Processors suitable for the execution of computer program include, by way of example, both general and special purpose microprocessors, and any one or more processor of any kind of digital computer. Generally, a processor will receive instructions and data from a read-only memory or a random access memory or both. The essential elements of computer are a processor for executing instructions and one or more memory devices for storing instructions and data. Generally, a computer will also include, or be operatively coupled to receive data from or transfer data to, or both, one or more mass storage devices for storing data, e.g., magnetic, magneto-optical disks, or optical disks. Information carriers suitable for embodying computer program instructions and data include all forms of non-volatile memory, including by way of example semiconductor memory devices, (e.g., EPROM, EEPROM, solid state drive (SSD), and flash memory devices); magnetic disks, (e.g., internal hard disks or removable disks); magneto-optical disks; and optical disks (e.g., CD and DVD disks). The processor and the memory can be supplemented by, or incorporated in, special purpose logic circuitry.

To provide for interaction with a user, the subject matter described herein can be implemented on a computer having an I/O device, e.g., a CRT, LCD, LED, or projection device for displaying information to the user and an input or output device such as a keyboard and a pointing device, (e.g., a mouse or a trackball), by which the user can provide input to the computer. Other kinds of devices can be used to provide for interaction with a user as well. For example, feedback provided to the user can be any form of sensory feedback, (e.g., visual feedback, auditory feedback, or tactile feedback), and input from the user can be received in any form, including acoustic, speech, or tactile input.

The subject matter described herein can be implemented in a computing system that includes a back-end component (e.g., a data server **413**), a middleware component (e.g., an application server), or a front-end component (e.g., a client computer **449** having a graphical user interface **454** or a web browser through which a user can interact with an implementation of the subject matter described herein), or any combination of such back-end, middleware, and front-end components. The components of the system can be interconnected through network **409** by any form or medium of digital data communication, e.g., a communication network. Examples of communication networks include cell network (e.g., 3G or 4G), a local area network (LAN), and a wide area network (WAN), e.g., the Internet.

The subject matter described herein can be implemented as one or more computer program products, such as one or more computer programs tangibly embodied in an information carrier (e.g., in a non-transitory computer-readable medium) for execution by, or to control the operation of, data processing apparatus (e.g., a programmable processor, a computer, or multiple computers). A computer program (also known as a program, software, software application, app, macro, or code) can be written in any form of programming language, including compiled or interpreted languages (e.g., C, C++, Perl), and it can be deployed in any form, including as a stand-alone program or as a module, component, subroutine, or other unit suitable for use in a computing environment. Systems and methods of the invention can include instructions written in any suitable programming language known in the art, including, without limitation, C, C++, Perl, Java, ActiveX, HTML5, Visual Basic, or JavaScript.

A computer program does not necessarily correspond to a file. A program can be stored in a portion of file **417** that holds other programs or data, in a single file dedicated to the program in question, or in multiple coordinated files (e.g., files that store one or more modules, sub-programs, or portions of code). A computer program can be deployed to be executed on one computer or on multiple computers at one site or distributed across multiple sites and interconnected by a communication network.

A file can be a digital file, for example, stored on a hard drive, SSD, CD, or other tangible, non-transitory medium. A file can be sent from one device to another over network **409** (e.g., as packets being sent from a server to a client, for example, through a Network Interface Card, modem, wireless card, or similar).

Writing a file according to the invention involves transforming a tangible, non-transitory computer-readable medium, for example, by adding, removing, or rearranging particles (e.g., with a net charge or dipole moment into patterns of magnetization by read/write heads), the patterns then representing new collocations of information about objective physical phenomena desired by, and useful to, the user. In some embodiments, writing involves a physical transformation of material in tangible, non-transitory computer readable media (e.g., with certain optical properties so that optical read/write devices can then read the new and useful collocation of information, e.g., burning a CD-ROM). In some embodiments, writing a file includes transforming a physical flash memory apparatus such as NAND flash memory device and storing information by transforming physical elements in an array of memory cells made from floating-gate transistors. Methods of writing a file are well-known in the art and, for example, can be invoked manually or automatically by a program or by a save command from software or a write command from a programming language.

Incorporation by Reference

References and citations to other documents, such as patents, patent applications, patent publications, journals, books, papers, web contents, have been made throughout this disclosure. All such documents are hereby incorporated herein by reference in their entirety for all purposes.

Equivalents

Various modifications of the invention and many further embodiments thereof, in addition to those shown and described herein, will become apparent to those skilled in the art from the full contents of this document, including references to the scientific and patent literature cited herein. The subject matter herein contains important information, exemplification and guidance that can be adapted to the practice of this invention in its various embodiments and equivalents thereof.

What is claimed is:

1. A nontransitory computer-readable medium storing software code representing instructions that when executed by a computing system cause the computing system to perform a method of reducing noise in a tomographic image, the method comprising

obtaining an A-scan from an imaging data set, the A-scan having a signal and the signal defining an amplitude;

estimating a noise floor specific to the A-scan by performing a moving average across the A-scan to obtain a miminum average wherein the mininmum average comprises the estimated noise floor;

scaling the amplitude of the signal based on the estimated noise floor;

combining the scaled A-scans to form a B-scan; and

scan-converting the B-scan into a Cartesian coordinate system to create a final tomographic view of the image with reduced noise.

2. The computer-readable medium of claim **1**, wherein the step of scaling comprises reducing the amplitude of the signal based on the estimated noise floor.

3. The computer-readable medium of claim **1**, wherein the moving average comprises computing the arithmetic mean, geometric mean, harmonic mean or quadratic mean within a sliding window across the A-scan.

4. The computer-readable medium of claim **1**, wherein the step of scaling further comprises adding a constant noise floor to the amplitude of the signal.

5. A nontransitory computer-readable medium storing software code representing instructions that when executed by a computing system cause the computing system to perform a method of reducing noise in a tomographic image, the method comprising

obtaining a plurality of A-scans from an imaging data set, each of the plurality of A-scans having a signal and the signal defining an amplitude;

estimating a noise floor specific to each of the plurality of A-scans;

scaling the amplitude of the signal in each of the plurality of A-scans based on the estimated specific noise floor for the corresponding A-scan;

combining the scaled A-scans to form a B-scan; and scan-converting the B-scan into a Cartesian coordinate system to create a final tomographic view of the image with reduced noise.

6. The computer-readable system of claim **5**, wherein the step of estimating comprises

averaging the amplitude of the signal at a plurality of pixels within a window of the A-scan.

7. The computer-readable medium of claim **5**, wherein the step of estimating comprises

performing a moving average across the A-scan.

8. The computer-readable medium of claim **7**, wherein the minimum average obtained from the performed moving average comprises the estimated noise floor.

9. The computer-readable medium of claim **5**, wherein scaling comprises reducing the amplitude of the signal based on the estimated specific noise floor.

10. The computer-readable of claim **6**, wherein the step of averaging comprises computing the arithmetic mean, geometric mean, harmonic mean or quadratic mean.

11. The computer-readable of claim **7**, wherein the moving average comprises computing the arithmetic mean, geometric mean, harmonic mean or quadratic mean within a sliding window across the A-scan.

12. The computer-readable medium of claim **5**, wherein the step of scaling further comprises adding a constant noise floor to the amplitude of the signal.

13. A system for imaging and reducing noise within a tomographic image, comprising:

a central processing unit (CPU); and

a storage device coupled to the CPU and having stored there information for configuring the CPU to:

obtain an A-scan from an imaging data set, the A-scan having a signal and the signal defining an amplitude;

estimate a noise floor specific to the A-scan by computing an arithmetic mean, geometric mean, harmonic mean or quadratic mean within a sliding window across the A-scan;

scale the amplitude of the signal based on the estimated noise;

combining the scaled A-scans to form a B-scan; and

scan-converting the B-scan into a Cartesian coordinate system to create a final tomographic view of the image with reduced noise.

14. The system of claim **13**, wherein the minimum average obtained from the performed moving average comprises the estimated noise floor.

15. The system of claim **13**, wherein scaling comprises reducing the amplitude of the signal based on the estimated noise floor.

16. The system of claim **13**, wherein scaling further comprises adding a constant noise floor to the amplitude of the signal.

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